Universidade de Lisboa

Faculdade de Motricidade Humana

Biomechanics of the trunk and lower limbs during gait in individuals with and without chronic low back pain

Tese elaborada com vista à obtenção do Grau de Doutor em Motricidade Humana na Especialidade de Biomecânica

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Julho de 2016

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The mere formulation of a problem is often far more essential than its solution, which may be merely a matter of mathematically or experimental skill. To raise new questions, new possibilities, to regard old problems from a new angle require creative imagination and marks real advances in science.

Albert Einstein

Dedicatória

Para a minha Mãe e para os meus sobrinhos Leonor, Mariana, Valentina e Mateus.

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Resumo

Integrar a informação sobre a cinemática e a cinética do tronco durante a marcha constitui um enorme desafio para as áreas da investigação e da clínica. Esta abordagem permite o aprofundamento do conhecimento sobre os mecanismos subjacentes aos padrões de movimento que se encontram alterados. Não obstante o crescente recurso à análise tridimensional da marcha para a avaliação de indivíduos saudáveis e com dor lombar crónica, a fiabilidade e o erro padrão de medida desta técnica não são ainda totalmente conhecidos. A presente dissertação procura resolver esta limitação, através do estudo das diferenças na biomecânica do tronco e dos membros inferiores durante a marcha em indivíduos saudáveis e com dor lombar crónica. Para a concretização desta dissertação, foram desenvolvidos três estudos. Os dois primeiros, com um desenho prospetivo, foram centrados na avaliação da fiabilidade e do erro de medição na análise tridimensional da marcha. Nestes estudos, os participantes (indivíduos saudáveis e com dor lombar crónica) foram submetidos a um protocolo de avaliação da marcha, com dois momentos distintos e com um intervalo médio de uma semana. Os dados foram recolhidos através de um sistema optoeletrónico (composto por treze câmaras) e de três plataformas de força. O processamento dos dados centrou-se nos parâmetros espaço-temporais da marcha, assim como nos valores máximos e mínimos dos ângulos e momentos articulares do tronco e membros inferiores. No terceiro estudo, com um desenho transversal, avaliaram-se as diferenças na biomecânica do tronco durante a marcha entre indivíduos saudáveis e indivíduos com dor lombar crónica. Para o efeito, determinou-se a variabilidade do movimento segmentar do tórax, lombar e anca e avaliou-se a sua correlação recíproca, tendo sido confirmada a presença de uma associação significativa entre as rotações residuais nos indivíduos com dor lombar crónica. Os estudos de reprodutibilidade revelam que a análise tridimensional da marcha é consistente, mas demonstram a presença de diferenças importantes na fiabilidade entre ângulos, momentos articulares e parâmetros espaço temporais, sendo o nível de erro todavia aceitável, sobretudo no plano sagital. Demonstrou-se ainda que a dor lombar crónica altera a variabilidade do movimento dos segmentos lombar e torácico durante a marcha e reduz a magnitude dos momentos articulares do tronco, também durante a marcha. Podemos, pois, afirmar que as alterações cinemáticas e cinéticas descritas suportam a existência de um padrão de proteção nestes indivíduos.

Palavras-Chave:

Fiabilidade; Erro de Medição; Coordenação Tronco-Pélvis; Dor Lombar Crónica.

Abstract

Combining information on kinetics and kinematics of the trunk during gait is important for both clinical and research purposes, since it can help in better understanding the mechanisms behind changes in movement patterns in chronic low back pain patients. Although three-dimensional gait analysis has been used to evaluate chronic low back pain and healthy individuals, the reliability and measurement error of this procedure have not been fully established. The main purpose of this thesis is to gain a better understanding about the differences in the biomechanics of the trunk and lower limbs during gait, in patients and healthy individuals. To achieve these aims, three studies were developed. The first two, adopted a prospective design and focused on the reliability and measurement error of gait analysis. In these test-retest studies, chronic low back pain and healthy individuals were submitted to a gait assessment protocol, with two distinct evaluation moments, separated by one week. Gait data was collected using a 13-camera opto-electronic system and three force platforms. Data analysis included the computation of time-distance parameters, as well as the peak values for lower limb and trunk joint angles/moments. The third study followed a cross sectional design, where gait in chronic low back pain individuals was compared with matched controls. Step-to-step variability of the thoracic, lumbar and hips was calculated, and step-to-step deviations of these segments from their average pattern (residual rotations) were correlated to each other. The reliability studies in this thesis show that three-dimensional gait analysis is a reliable and consistent procedure for both chronic low back pain and healthy individuals. The results suggest varied reliability indices for multi-segment trunk joint angles, joint moments and time-distance parameters during gait, together with an acceptable level of error (particularly regarding sagittal plane). Our findings also show altered stride-to-stride variability of lumbar and thoracic segments and lower trunk joint moments in patients. These kinematic and kinetic results lend support to the notion that chronic low back pain individuals exhibit a protective movement strategy.

Keywords

Reliability; Measurement Error; Trunk-Pelvis Coordination; Chronic Low Back Pain.

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List of Abbreviations

2D - Two-dimensional

3D - Three-dimensional

3DGA - Three-dimensional gait analysis

ASIS - Anterior superior iliac spine

BMI - Body mass index

BPAQ - Baecke physical activity questionnaire

CI – Confidence interval

CLBP - Chronic low back pain

D - Mean difference between measurements

GBD - Global burden of disease

GCVSPL – Generalized cross-validatory cubic spline smoothing routine

ICC - Intraclass correlation coefficient

LBP – Low back pain

LCS - Local coordinate system

LOA - Limits of agreement

MDC - Minimal detectable change

MSE – Mean standard error

NRS - Numerical rating scale

OR - Odds ratio

PSIS - Posterior superior iliac spine

QBPDS - Quebec back pain disability scale

r_s - Spearman correlation coefficients

R_s^2 - Correlation coefficients squared

SD_{diff} - Standard deviation of the difference

SE – Standard error

SEM - Standard error of measurement

SO - Segment optimization

STA - Soft tissue artifacts

TSK - Tampa scale of kinesiophobia

YLDs - Years lived with disabilities

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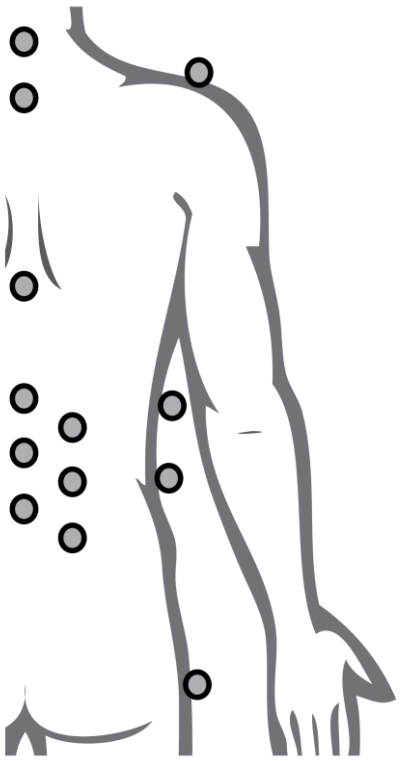
General Introduction



General Introduction

Low back pain (LBP) is one of the most common health problems in society and causes considerable disability, work absenteeism, and use of health services (Cassidy, Côté, Carroll, & Kristman, 2005). A systematic analysis for the Global Burden of Disease (GBD) published in the Lancet (Vos et al., 2012) reported that LBP stands out as the leading musculoskeletal disorder because of a combination of high prevalence and greater disability weight associated with this health state. According to this report, LBP is one of the four most common disorders in all regions, and is the leading cause of years lived with disabilities (YLDs) in all developed countries. Low back and neck pain account for 70% of all YLDs from musculoskeletal disorders, and for every YLD due to neck pain there are 2.5 YLDs related to LBP. The burden as estimated in this report is substantially higher than previously assessed in the GBD 1990 and GBD 2000 rounds of estimations, which combined with the 33.3% increase in YLDs from 1990 to 2010 driven largely by population growth and ageing, have important implications for health systems. Estimates from GBD also identify low back and neck pain as the primary cause of YLDs in Portugal, with an increase of 13% from 1999 to 2013 (Burden global of Disease, 2013).

A meta-analysis on the clinical course of pain and disability in patients with acute LBP, confirm the earlier findings that typical course of acute LBP is initially favourable, i.e. is characterized by a marked reduction in the mean of pain and disability in the first six weeks (Costa et al., 2012). However, from this point forward, improvement slows down and thereafter only small reductions in mean pain and disability are apparent up to one year, when low to moderate levels of pain and disability are expected. A systematic review on the long-term course of LBP indicates that after an episode of low back pain, 42 to 75% of the patients still experience pain 12 months later (Hestbaek, Leboeuf-Yde, & Manniche, 2003). Moreover, 44% to 78% suffer from a relapse of back pain, 26 to 37% have relapses of work absence and 26% to 37% suffer from recurrent sick leave. The study also reports that the prevalence of LBP in cases with previous episodes is 56% (range 14–93%), compared with 22% (range 7–39%) for those without a prior history of LBP.

Based on time-related criteria, LBP can be classified as acute (up to six weeks), sub-acute (between six weeks and three months) and chronic (over three months) (Koes, van Tulder, & Thomas, 2006). Especially chronic low back pain (CLBP) has a large economic impact, mainly because of the above described sickness absence and long-

term disability and is considered to be one of the largest health related challenges in industrialized societies (Parthan, Evans, & Le, 2006). In Portugal, epidemiological findings have shown that musculoskeletal chronic pain is one of the most common complaints in general population (36.7%), with CLBP being among the main reasons for patients' sick health care (Azevedo, Costa-Pereira, Mendonça, Dias, & Castro-Lopes, 2012; Castro-Lopes, Saramago, Romão, & Paiva, 2010). A recent study conducted under the scope of EpiReumaPt found that among 10.661 subjects, 1487 self-reported CLBP, resulting in a prevalence of 10.4 % (95 % CI, 9.6 to 11.9 %) (Gouveia et al., 2016). This study also found that CLBP was associated with disability and with a high use of healthcare resources.

In the absence of a specific patho-anatomic diagnosis, approximately 90% of the individuals with LBP are labelled as “non-specific low back pain” or some equivalent term (Hancock, Maher, Laslett, Hay, & Koes, 2011) which, in essence, is a diagnosis based on exclusion of specific pathology (Koes et al., 2006). Consequently, a generic symptomatic treatment is applied and the results obtained are not satisfactory. A great variety of interventions, including multidisciplinary treatment, cognitive behavioural therapy and supervised exercise therapy, have been proposed for the treatment of non-specific CLBP (Airaksinen et al., 2006; Koes et al., 2010). However, the available evidence from placebo-controlled trials shows only small to moderate analgesic treatment effects, over and above placebo, for many interventions that are currently used in the management of non-specific acute or chronic LBP (Machado, Kamper, Herbert, Maher, & McAuley, 2009). The limitations of current approaches are further illustrated by the many systematic reviews that reveal that existing treatments for non-specific CLBP have, at best, only small effects (Deyo, 2004; Machado et al., 2009). One example is a systematic review that assessed the overall responses to treatments among non-specific LBP patients in 118 clinical trials (Artus, van der Windt, Jordan, & Hay, 2010). Results showed a similar pattern of initial improvement at 6 weeks followed by smaller improvement for both pain and functional disability at long-term follow-up. This was also shown by the pooled standardized mean difference for pain, which was 0.86 (95% CI, 0.65 to 1.07) at 6 weeks, 1.07 (95% CI, 0.87 to 1.27) at 13 weeks, 1.03 (95% CI, 0.82 to 1.25) at 27 weeks and 0.88 (95% CI, 0.60 to 1.1) at 52 weeks. The rational behind the modest effect of current interventions is not clear, however, one possible explanation is the heterogeneity in development of persistent pain trajectories between patients. Studies show that the course can differ per individual or group: some improve more rapidly, some more slowly, whereas others may fluctuate (Dunn, Jordan,

& Croft, 2006). The modest benefit may also be explained by patients' response to treatment, since it can be assumed that individuals with different characteristics may respond differently to specific rehabilitation programmes.

Considering the high prevalence rates of CLBP and the limitations of current approaches, it is important to gain insight regarding the factors that may be associated with the development and course of this condition, as well as the ones influencing its outcome prognosis. This information is particularly important for further development of specific interventions based on previously identified modifiable prognostic factors. Literature has shown different groups of factors that can contribute to CLBP and disability, including psychosocial and biological ones (Wand & O'Connell, 2008). Psychological factors are an important part of the chronic LBP experience. They contribute to chronicity, explain a significant amount of the variance of different outcomes (e.g. disability, pain or return to work) among CLBP patients and have been identified as important barriers to pain resolution (Linton, 2000; Pincus, Burton, Vogel, & Field, 2002; Pincus, Vogel, Burton, Santos, & Field, 2006). Maladaptive coping strategies such as negative thinking, pathological fear and abnormal anxiety regarding pain, avoidant behaviour, catastrophizing and hypervigilance have been shown to be associated with high levels of pain, disability and muscle guarding (Linton, 2000; Main & Watson, 1996; Nachemson, 1999). Social factors such as the compensation system, workplace disputes and cultural issues affecting beliefs reinforce the psychological factors that can increase pain (Nachemson, 1999).These factors have been extensively studied and identified as key for the development and course of pain and disability in these patients. However, above described psychosocial factors seem to fall short as prognostic factors for recovery in CLBP. For instance, they seem to be of non-importance in a prognostic model on absolute and relative (30%) recovery on pain and disability in CLBP patients. In this study, at 5-month follow-up the prognostic factor most strongly associated with relative recovery in pain is Body Mass Index (BMI) of ≥ 25-29.9 kg/m^2 (Odds Ratio (OR) 1.27 95% CI, 0.99 to 1.62) and higher work participation at baseline (OR 1.27 95% CI, 0.93 to 1.73), while at 12-month follow-up is being married or living with one adult (OR 1.6 95% CI, 0.99 to 2.57) (Verkerk et al., 2015). Work participation (OR 1.34 95% CI, 0.93 to 1.93) (5-month follow-up), back pain intensity in the previous 3 months (OR 1.42 95% CI, 1.02 to 1.99) and BMI ≥ 30 kg/m^2 (OR 1.74 95% CI, 1.10 to 2.76) (12-month follow-up), are the strongest prognostic factors for absolute recovery (Verkerk et al., 2015). Factors of importance for recovery in disability at 5 and 12 months are younger age and higher scores on

disability and on the 36-Item Short-Form Health Survey at baseline. At 5-month follow-up, a shorter duration of complaints is a positive predictor, and having no comorbidity and less pain at baseline are additional predictors at 12-month follow-up (Verkerk & Luijsterburg, 2013).

Besides psychosocial factors, also biological ones are described to contribute to the development, persistence and recurrence of LBP. These factors are broader than potential nociceptive sources and include central modulation of pain and physical impairments (Hancock et al., 2011). Regarding the latter, people with LBP show notable limitations in both spinal and hip motion that compromise function, which may have impact on their quality of life (Shum, Crosbie, & Lee, 2005a). Studies that attempted to estimate these mobility impairments have focused on different functional activities and identified altered biomechanical patterns in LBP individuals during sit-to-stand and reverse (Shum et al., 2005a), putting on a sock (Shum, Crosbie, and Lee 2005b) and backward/forward bending (Shum, Crosbie, & Lee, 2010). In CLBP, motor control tasks seem to be altered, with patients demonstrating loss of variability during functional tasks and delayed reflexes (van Dieën, Cholewicki, & Radebold, 2003). This might also interfere with walking. Many CLBP individuals report problems with this complex activity, which is probably the reason why it has been the focus of studies, particularly with respect to its kinematics characteristics. Based on prior research, it appears that people with CLBP who are allowed to self-select walking speed, consistently walk slower (Lamoth et al., 2002; Lamoth, Stins, Pont, Kerckhoff, & Beek, 2008; Müller, Ertelt, & Blickhan, 2015), take shorter steps and have asymmetric step lengths, when compared with their healthy peers (Keefe & Hill, 1985; Vogt, Pfeifer, Portscher And, & Banzer, 2001). Regarding joint angles, literature shows conflicting results: some investigators have reported that people with CLBP display less axial rotation of the pelvis (Müller et al., 2015) or the lumbar segment (Gombatto et al., 2015), while others found no significant differences in absolute axial rotation of the trunk, thorax or pelvis between CLBP individuals and controls (Lamoth et al., 2002; Seay, Van Emmerik, & Hamill, 2011). Furthermore, one study reported that the degree of lumbar or thorax axial rotation depended on temporal parameters of the gait cycle (Crosbie, de Faria Negrão Filho, Nascimento, & Ferreira, 2013). Specifically, the authors verified that comparatively to healthy controls, recurrent LBP patients show less axial rotation of the thorax and lumbar during mid-stance and heel strike, respectively, but higher axial rotation of the lumbar during mid-stance. According to literature, the mentioned changes in trunk and pelvis mobility during gait in CLBP

individuals do not seem to be responsible for the development of LBP itself (Hodges & Tucker, 2011). They play a major role as an adaptive response to allow a short-term protection from further pain, injury, or both. This adaptation may have consequences that could lead to further problems in the long term (Hodges & Moseley, 2003; Hodges & Tucker, 2011).

To study the adaptive response to LBP, many researchers looked at different responses in variability and coordination of the trunk and pelvis during gait. Recent findings suggest that patients exhibit a reduced ability to adapt trunk–pelvis coordination in response to changes in gait velocity (Lamoth, Daffertshofer, Meijer, & Beek, 2006), display a more rigid and less flexible pelvis-thorax coordination (Lamoth et al., 2002), and have lower variability of trunk rotations, as a result of the coupling of deviations of residual rotations (in shape and amplitude) between pelvis and trunk (van den Hoorn, Bruijn, Meijer, Hodges, & van Dieën, 2012). Crosbie et al. (2013) also suggest that coordination between adjacent segments might be dependent of gait cycle phase, i.e. limited motion of the pelvis translates to reduced lumbar and lower thoracic angular displacement in LBP individuals at mid-stance sub-phase of the gait cycle. These findings are in disagreement with Vogt et al., (2001) who verified higher stride-to-stride variability and increased fluctuations in dynamic oscillations of angular displacement of thoracic and pelvic segments, in a sample of CLBP patients. These contradictory results may be explained by the fact that, in these studies, different methods were used to compute kinematics variability (Crosbie et al., 2013; Lamoth et al., 2006; van den Hoorn et al., 2012; Vogt et al., 2001). Moreover, the use of different walking surfaces (overground vs. treadmill), biomechanical models and the heterogeneous nature of LBP, may also have contributed to the conflicting results, even though it was suggested that these measures provide valuable information when assessing the quality of gait in these patients (Lamoth et al., 2002).

The kinematic analysis of functional activities is highly valuable. However, it remains descriptive and cannot fully explore the biomechanical mechanisms underlying changes in movement strategies and the nature of the loading patterns in the lumbar spine (Shum, Crosbie, & Lee, 2007). Previous studies that attempted to estimate kinetic variables in LBP individuals have mainly focused on functional activities that included flexion and extension of the trunk, namely lifting tasks (Kingma et al., 2001; Marras, Davis, Ferguson, Lucas, & Gupta, 2001), sit-to-stand and reverse (Shum, Crosbie, & Lee, 2009; Shum et al., 2007), as well as backward/forward bending (Shum, Crosbie, & Lee, 2010). Those studies found that, compared to healthy subjects, LBP

individuals had decreased sagittal joint moments acting on the lumbar spine at the end of the available range during forward/backward bending and sit-to-stand, had increased transverse plane joint moments during sit-to-stand. In addition, a decreased muscle power around the lumbar spine and hip was demonstrated in LBP individuals during sit-to-stand and stand-to-sit. This, as well as the above mentioned changes in kinematics and functional tasks, contribute to an explanatory theory that patients adopt a protective strategy in terms of reduced joint moments and powers acting on the spine and hips, in order to prevent further pain (Shum et al., 2007, 2010). In line with this hypothesis, previous studies also showed that LBP individuals recruit their muscles differently and have alterations of the flexion-relaxation response typically seen in asymptomatic counterparts (Alschuler, Neblett, Wiggert, Haig, & Geisser, 2009). Specifically, in LBP individuals flexion-relaxation was absent or significantly impaired (surface electromyographic activity persists at full trunk flexion) (McGorry & Lin, 2012), suggesting distinctive muscle activation patterns that may impose an altered load on the lumbar spine (Shum et al., 2007). According to our best knowledge, studies with CLBP individuals that focused on complex activities, such as gait, have limited their analysis to kinematic and electromyographic variables (Gombatto et al., 2015; Lamoth et al., 2006; van den Hoorn et al., 2012; Vogt et al., 2001). Combining information on the kinetics and kinematics of the trunk during gait is of importance, since it can help in better understanding the mechanisms behind the changes in movement patterns in CLBP.

Currently, three-dimensional (3D) analysis is considered a valid measurement tool to study change in motor adaptive patterns during gait and is classified as the 'gold standard' method (Meldrum, Shouldice, Conroy, Jones, & Forward, 2014). Validity refers to whether a given instrument or test measures what it aims to measure (Streiner, Norman, & Cairney, 2015). Another essential requirement of all outcome measures is its reproducibility. This concerns to the degree to which repeated measurements in stable study subjects provide similar results, and depends on the measurement error (how close the scores for repeated measurements are) and reliability (how well can subjects be distinguished from each other, despite measurement errors) (de Vet, Terwee, Knol, & Bouter, 2006). Information regarding the reproducibility of three-dimensional gait analysis (3DGA) still requires more investigation within healthy and clinical populations. Comparisons with asymptomatic participants have been made, profiles have been suggested and, in some cases, recommendations regarding intervention strategies were made, without taking into

account the measurement error and the minimal detectable change (MDC) of the measured gait variables.

Variability between 'before' and 'after' a given intervention may be due to treatment effects or measurement variation, or a combination of both. Knowledge on the error magnitude can enable clinical teams to minimise the risk of over-interpreting small differences as meaningful (Schwartz, Trost, & Wervey, 2004) and to have greater confidence that the treatment effect exceeds the measurement error. Additionally, the use of measurements with low reliability in clinical research may lead to underestimation or failure to detect significant effect sizes; with too much noise (error) drowning out real effects (McGinley, Baker, Wolfe, & Morris, 2009). To address change accurately in health-related outcomes, clinicians need measurement tools that show responsiveness and are able to detect minimal changes in performance over time (Streiner et al., 2015). This change must be large enough to be considered a "real" change and precise enough to detect small but important clinical changes over time considered to be important by patients and/or clinicians (Demoulin, Ostelo, Knottnerus, & Smeets, 2010; Terwee et al., 2007). The observed variability in gait data can be attributed to two sources: intrinsic variability or true variation in the patient's gait pattern, and extrinsic variability due to methodological errors in marker placement, anthropometric measurements, or calibration of the motion capture system. Another source of variability in 3DGA is the "soft tissue artifacts" (STA), that arises from movement or deformation of the subcutaneous tissues associated with muscular contractions, skin movement and inertial effects (Cappozzo, Catani, Leardini, Benedetti, & Croce, 1996). The extent of STA for any movement depends upon the physical characteristics of individuals, marker locations and the nature of the movement task being performed (Peters, Galna, Sangeux, Morris, & Baker, 2010). While intrinsic variation cannot be reduced, the measurement variation that arises from extrinsic factors can be controlled (Schwartz et al., 2004). It is generally accepted that two major sources of error in 3DGA data are marker placement and STA, although other factors, such as inconsistent anthropometric measurements, variation in walking speed, data processing or measurement equipment errors, may also contribute to data variation (Monaghan, Delahunt, & Caulfield, 2007). Although 3DGA constitutes a complex procedure for daily clinic, the ability of clinicians to discern findings that are meaningful from those that are insignificant or artifactual is nonetheless essential.

A number of studies have evaluated the reliability of kinematic and kinetic parameters during gait in healthy and clinical populations, namely cerebral palsy, stroke,

adolescent idiopathic scoliosis, hip osteoarthritis, cervical spondylotic myelopathy and incomplete spinal cord injury. Approaches to the analysis of reliability in 3DGA have differed among studies. Some examined the reliability of kinematic and kinetic curves over the complete gait cycle (Delval et al., 2008; Meldrum et al., 2014; Schwartz et al., 2004; Steinwender et al., 2000), while others extracted key points from those curves, such as a peak value or a range (Fortin, Nadeau, & Labelle, 2008; Klejman, Andrysek, Dupuis, & Wright, 2010; McDermott, Bolger, Keating, McEvoy, & Meldrum, 2010). Key points have been considered more meaningful, as they are easier to compare and interpret than complete curves, and tend to include the most clinically relevant features of the curves (Redekop, Andrysek, & Wright, 2008). A systematic review based on the results of 15 studies using 3DGA, found highest reliability for kinematic parameters in the sagittal plane (intraclass correlation coefficient (ICC) 0.8), with the exception of pelvic tilt (ICC 0.6), and lowest reliability in the transverse plane (ICC <0.7) (McGinley et al., 2009). The authors reported standard error of measurement (SEM) values around 4º in the sagittal plane and 2º in the frontal plane, concluding that most kinematic parameters showed moderate to good reliability, but not small enough measurement errors that may be ignored in clinical interpretation. So, to study adaptive responses in CLBP during 3DGA, reliability and measurement error analysis have to be incorporated.

Although there is evidence that clinically acceptable errors are possible in 3DGA in both healthy individuals and patients (e.g. cerebral palsy or stroke), data on the reliability and measurement error of 3DGA in CLBP patients is lacking. According to a systematic review (Mieritz, Bronfort, Kawchuk, Breen, & Hartvigsen, 2012), studies that evaluated the reproducibility of 3D spinal motion analysis in CLBP patients focused on simple movements (e.g. flexion or extension of the trunk) and are difficult to interpret due to incomplete reporting of the studies' populations, testing protocol, statistics and data presentation. The majority of the included studies did not report agreement parameters, which may question the performance of the evaluated instruments in clinical practice. Recent studies that aimed to examine the reliability and measurement error of 3D spinal motion parameters, verified that sagittal and frontal plane kinematics of the trunk (modelled as a whole segment) may be sufficiently reliable in measurements of groups of CLBP patients (Harsted, Mieritz, Bronfort, & Hartvigsen, 2016; Mieritz, Bronfort, Jakobsen, Aagaard, & Hartvigsen, 2014). Since reliability of measurement tools can be population (Streiner & Norman, 2008) and task specific, studies with the purpose of investigating test-retest reliability and MDC of 3DGA in a sample of CLBP patients are needed.

Additionally, to assess change in walking over time, clinical gait analysis typically seeks to compare between normal and abnormal gait (McGinley et al., 2009). Moreover, knowledge about reliability and MDC values from healthy population is extremely important since it can help clinicians and researchers interpret pathological data. As previously mentioned, several studies have investigated the reliability of 3DGA in healthy individuals and patients, revealing error values of less than 5º for all gait variables, excluding hip and knee rotation (McGinley et al., 2009). Likewise, moderate to good reliability for sagittal and frontal plane variables was reported, with the exception of pelvic tilt and knee varus/valgus in some studies. Two studies (Meldrum et al., 2014; Wilken, Rodriguez, Brawner, & Darter, 2012) provided absolute measures of measurement error and MDC values for kinematic and kinetic parameters in healthy individuals. Meldrum et al. (2014) reported low SEM (≤5˚) for the majority of the lower limb kinematic parameters and variable ICC values (0.14 to 0.92), while Wilken et al., (2012) reported good to excellent reliability of lower limb and trunk kinematics/kinetics across a range of controlled walking velocities, and low MDC values (approximately of 5º for joint angles). By adding trunk data, Wilken's study made an important contribution to the knowledge on this topic. However, the authors considered the trunk as one rigid segment and excluded information regarding transverse plane kinetics, which may contribute to clinical reasoning and decision-making when dealing with movement disorders. Considering that sufficient evidence exists supporting that different regions of the trunk move differently, we can argue that one of the main limitations of the prior work is that the whole trunk was considered a single rigid segment. Thus, studies aimed at investigating reliability and MDC of kinematics and kinetics of multi-segment trunk in 3DGA in healthy and CLBP individuals are needed.

Thesis Aims and Methodology Synopsis

Although 3DGA has been used to measure kinematics in CLBP individuals, the reliability and measurement error of this evaluation procedure, in this specific population, has not been established. Additionally, trunk kinematics and kinetics can contribute to more detailed information on gait impairment. However, data on reliability and measurement error of multi-segment trunk on 3DGA is lacking on both healthy and CLBP individuals. Once we have the information on reliability and agreement, it will be possible to rigorously compare the gait of individuals with and without CLBP and to gain insight into the differences between their movement patterns.

Based on the lacking or conflicting knowledge described and discussed throughout this chapter, the main purpose of this thesis is to gain a better understanding about the differences in the biomechanics of the trunk and lower limbs during gait in CLBP and healthy individuals. Specifically, the established aims are:

- To investigate test-retest reliability and MDC of 3DGA in a sample of CLBP patients.
- To investigate test-retest reliability and MDC of 3DGA kinematic and kinetic data in a sample of healthy individuals, using a two rigid segment model for the trunk.
- To compare lumbar and thoracic kinematics and kinetics between CLBP and healthy individuals during gait, taking into account the error values.
- To assess the variability of movement between lumbar and thoracic segments, in association with joint moments and angles, in CLBP patients versus healthy individuals.

With our two final goals we expect to have a better understanding on the differences between the movement patterns' of CLBP and healthy individuals, which will broaden future research and open up possibilities about how to change motor adaptive patterns and protective strategies in patients.

To achieve these aims, the thesis is divided into three distinct but related studies. In the first two, the focus lies on the reliability and measurement error studies. Two prospective (within assessor) test-retest studies, where participants (CLBP or healthy individuals) underwent two biomechanical gait assessments with a mean interval of one week, were conducted. Gait data were collected using a 13-camera opto-electronic system and three force platforms. Participants were instructed to walk during a few minutes at their preferred velocity and 10 gait cycles were selected for further processing. A Woltring generalized cross-validatory cubic spline smoothing routine was applied to kinematic and kinetic data. The marker set selection was based on previous reports (Leardini, Biagi, Merlo, Belvedere, & Benedetti, 2011; Seay, Selbie, & Hamill, 2008) and a 9 segments model (feet, shanks, thighs, pelvis, lumbar and thoracic segments) was built and optimized through segment optimization. Data analysis included ICCs, the mean difference between measurements (D), and the 95% Confidence interval (CI) for D, the standard deviation of the differences (SD_{diff}) and the 95% Bland and Altman limits of agreement (95% LOA) for anthropometric, time-distance and key kinematic/kinetic parameters. The third study followed a cross

sectional design, where gait of CLBP individuals was compared with matched controls. The data collection system, marker set, biomechanical model and laboratory procedures for data collection were the same as in the reliability studies. Filtering process and data optimization also followed the same procedures. Data analysis included the computation of time-distance parameters, peak values for hip and trunk joint angles/moments. Step-to-step variability of the thoracic, lumbar and hips was also calculated, and step-to-step deviations of these segments from their average pattern (residual rotations) were correlated to each other.

Thesis Outline

Chapter 2 presents a study in which reliability and minimal detectable change of 3DGA is tested in a sample of CLBP patients.

Chapter 3 reports the results of a study that evaluated the reliability and minimal detectable change of a two rigid segment model for the trunk during 3DGA, in sample of healthy individuals.

Chapter 4 describes the results of study that compared the variability of movement between lumbar and thoracic segments, in association with joint moments and angles, in CLBP patients versus healthy individuals. In this study the results of the lumbar and thoracic kinematics and kinetics are interpreted according to error values obtained in the reliability studies.

Chapter 5 addresses the main results of this thesis, discusses some methodological options, and makes some recommendations for further research.

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2

Test-retest reliability and minimal detectable change of three-dimensional gait analysis in chronic low back pain patients

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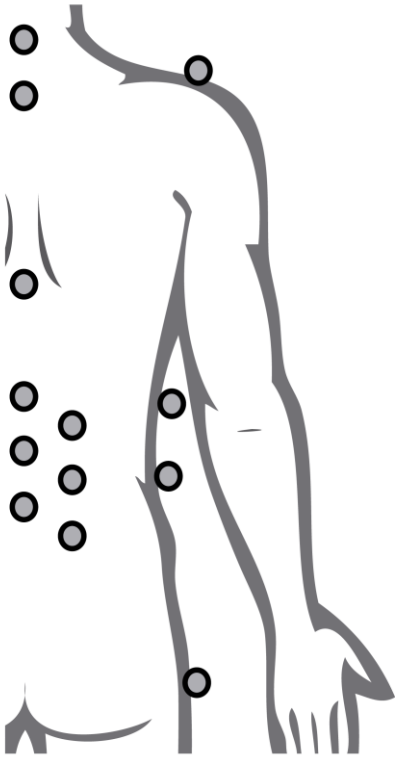


Abstract

Background and Aim: Three-dimensional gait analysis can provide detailed data on gait impairment in CLBP patients. However, data about reliability and measurement error of 3DGA in this population is lacking. The aim of this study is to investigate test-retest reliability and minimal detectable change of 3DGA in a sample of CLBP patients.

Methods: A test-retest study was conducted with a sample of 14 CLBP patients that underwent two biomechanical gait assessments with an interval of 7.6±1.8 days. Anthropometric and time-distance parameters, as well as peak values for lower limb and trunk joint angles and moments, were computed. $ICC_{3,k}$ and their 95% confidence intervals were calculated. SEM, MDC and limits of agreement (LOA) were also estimated.

Results: The obtained ICC values demonstrate high test-retest reliability for most joint angles, with low SEM (<2.5º) values. Although joint moments showed lower reliability than joint angles, the majority of the ICCs were above 0.7 and the SEM and MDC values were low (≤0.06 Nm/kg and ≤0.18 Nm/kg). Bland-Altman plots with 95% LOA revealed a good agreement and time-distance parameters were all highly repeatable (ICCs > 0.86).

Conclusions: The results of this study show high test-retest reliability for lower limb and trunk joint angles, and time-distance parameters during gait in CLBP individuals, together with a low measurement error. These results also support the use of this method in clinical assessments of CLBP patients' gait patterns.

Keywords

Gait analysis; Reliability; Measurement error; CLBP.

2.1 Introduction

Chronic Low back pain is a common health condition in western industrialised countries with an estimated prevalence of 20.1±9.8% (Hoy et al., 2012). Patients often report difficulties during daily activities, such as gait. Studies have reported that gait coordination is changed in CLBP patients: they walk slower, take shorter steps and have asymmetric step lengths when compared with their healthy peers (Keefe & Hill, 1985; Vogt, Pfeifer, Portscher And, & Banzer, 2001). Chronic low back pain patients also have difficulty in moving from pelvis-trunk in-phase to anti-phase (pelvis and trunk moving in the same or in opposite directions, respectively) as walking speed increases (Lamoth et al., 2002) and consequently show lower variability of trunk rotations, possibly adopting a protective movement strategy to diminish pain (Huang et al., 2011).

In clinical settings, gait evaluation in CLBP patients is frequently carried out by observation and functional tests (Andersson, Lin, & Smeets, 2010), or is included in specific disability questionnaires (Malliou, Gioftsidou, Beneka, & Godolias, 2006), which only provide limited information. In contrast, although time consuming, 3DGA can provide detailed quantitative data concerning gait impairment (Baker, 2013). As an advantage in CLBP patients, 3D instruments can obtain real-time information on 3D lumbar spine kinematics and kinetics without any known risk to the patients (Mieritz, Bronfort, Jakobsen, Aagaard, & Hartvigsen, 2014). Thus, 3DGA can assist in reaching clinical functional diagnoses and can be useful to evaluate the outcome of therapeutic interventions (Mieritz et al., 2014). However, as with any analysis tool, reliability and measurement error emerge as critical factors in its applicability to clinical decision-making (McDermott, Bolger, Keating, McEvoy, & Meldrum, 2010). Since low reliability in clinical research may lead to underestimation or failure to detect significant effect sizes (McGinley, Baker, Wolfe, & Morris, 2009), we have to strive for good reliability. In addition, knowledge of the error's magnitude can minimise the risk of over-interpreting small differences as meaningful (de Vet et al., 2006) and contribute to the certainty that a measured intervention effect exceeds the measurement error.

Data on reliability and measurement error of 3DGA in CLBP patients is lacking, although evidence exist that clinically acceptable errors are possible in 3DGA in healthy individuals and in patients with cerebral palsy or stroke (McGinley et al., 2009). The few studies that evaluated reliability and measurement error of 3D spinal motion analysis in CLBP patients (Mieritz et al., 2014; Mieritz, Bronfort, Kawchuk, Breen, & Hartvigsen, 2012) focused on simple activities and are difficult to interpret due to

incomplete reporting of the studies' populations, testing protocol, statistics and data presentation (Mieritz et al., 2012). Since reliability of measurement tools can be population (Streiner & Norman, 2008) and task specific, the aim of this study was to investigate test-retest reliability and minimal detectable change of 3DGA in a sample of CLBP patients.

2.2 Materials and Methods

2.2.1 Study Design

A prospective (within assessor) test-retest study was conducted.

2.2.2 Participants

A convenience sample of 23 CLBP patients was recruited from community and outpatient clinics according to a standardized recruitment protocol. Firstly, physiotherapists from the research team and outpatient clinics carried out patient recruitment based on predefined inclusion/exclusion criteria. Patients were considered eligible if they were aged between 18 and 65 years, and had LBP, with or without referred leg pain, for at least 12 weeks (Airaksinen et al., 2006) or recurrent LBP (Von Korff, 1994). Eligible patients were screened for evidence of serious low back pain pathology and were excluded if they had clinical signs of infection, tumour, osteoporosis, fracture, structural deformity, inflammatory disorder (e.g. ankylosing spondylitis), radicular syndrome, cauda equine syndrome, or if they had undergone back or lower limb surgery or a conservative treatment in the prior 12 and 6 months, respectively. Pregnant women were also excluded. After this screening, 14 of the 23 patients agreed to perform two consecutive assessments with a mean interval of 7 days.

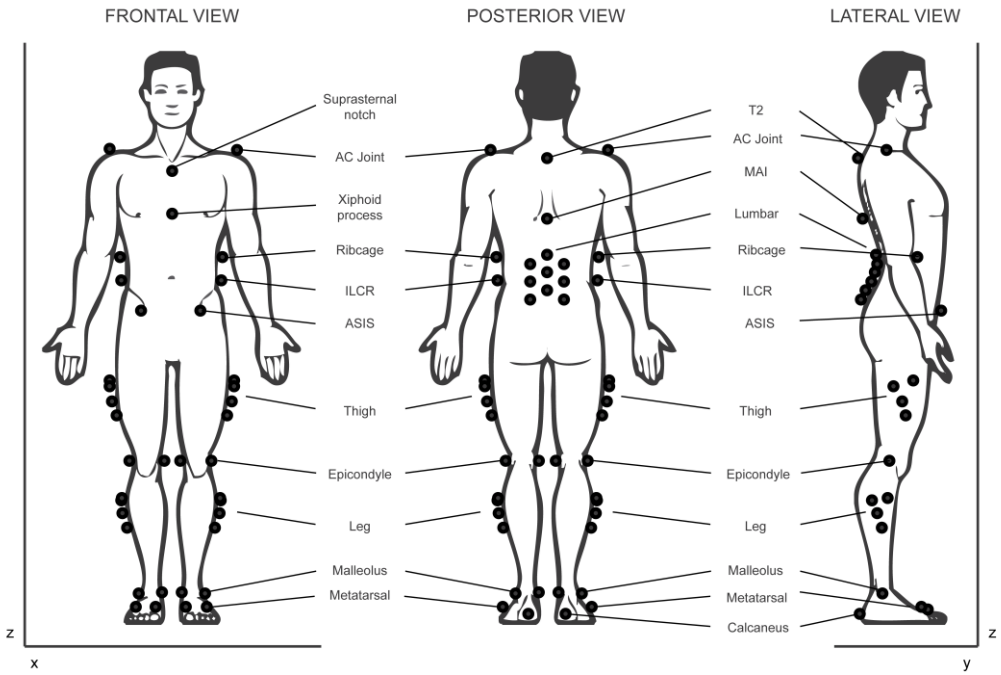
The local Ethics Committee approved the study. All the participants were informed of the procedures and risks of the study and signed an informed consent.

2.2.3 Procedures

Gait analysis was performed twice with an interval of 6 to 11 days (mean ± standard deviation, 7.6±1.8). This time interval was considered long enough to avoid assessor memory bias and short enough to avoid a change in patients' gait pattern or clinical condition (McDermott et al., 2010). On the first visit to the laboratory, participants' clinical history was reviewed and a standard physical examination focussed on lumbar spine and lower limbs was performed. This was complemented with the measurement

of body mass and height. Segments' length was obtained using the respective proximal and distal anatomical landmarks collected during the static trial described below. For pelvis, anterior and posterior superior iliac spine (ASIS and PSIS) markers were used. To assure participants' clinical stability between test and retest sessions, pain intensity and disability were assessed using the Numerical Rating Scale (NRS) and the Portuguese version of the Quebec Back Pain Disability Scale (QBPDS-PT), respectively. Details regarding the psychometric properties of these measurement tools can be found elsewhere (Cruz et al., 2013).

Finally, gait data was collected using a 13-camera opto-electronic system (Oqus 300, Qualisys AB, Gothenburg, Sweden) synchronized in time and space with two Kistler (9281B and 9283U014, Kistler Group, Winterthur, Switzerland) and one AMTI (BP6001200, Advanced Mechanical Technology, Inc Watertown, USA) force platforms, at 200 Hz. The marker set used was based on previous reports (Leardini, Biagi, Merlo, Belvedere, & Benedetti, 2011; Seay, Selbie, & Hamill, 2008) (Figure 1). After a static trial, participants were instructed to walk barefoot at their preferred velocity, continuously and during short periods of time (1-2 minutes). A familiarization trial was performed before data collection. Each participant was assessed at the same time of the day to minimize the effects of diurnal variations in joint mechanics. All the procedures were carried out by the same assessor.



Abbreviations: T2, Second thoracic vertebra; AC, Acromioclavicular; MAI, midpoint between the inferior angles of most caudal points of the two scapula; ILCR, Iliac crest; ASIS, Anterior superior iliac spine.

Figure 1 - Marker Set

2.2.4 Data Processing

Considering the natural variability in kinematic and kinetic gait parameters, 10 cycles were selected (Monaghan, Delahunt, & Caulfield, 2007). Cycles were extracted using Qualysis Track Manager (v2.8 build 1554, Qualisys AB, Gothenburg, Sweden) and exported to be processed under Visual 3D software (v5.01.10, C-Motion, Inc, Rockville, USA).

A 9-segment model (feet, legs, thighs, pelvis, lumbar and thoracic spine) was built for each participant (Leardini et al., 2011; Seay et al., 2008). Each segment was considered to be independent and to have 6 degrees of freedom (segment optimization (SO) method) (Cappello, La Palombara, & Leardini, 1996). Lower limb segment masses were determined according to Dempster (1955) while the remaining inertial parameters were computed based on Hanavan (1964). Lumbar and thoracic inertial parameters were computed according to Pearsall, Reid and Livingston (1996). The ankle and knee joint centres were defined as the midpoint of the tibia malleoli and as the midpoint of the femur epicondyles, respectively (Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2014). The hip joint centres were computed using the pelvis markers, according to published regression equations (Bell, Pedersen, & Brand, 1990). The lumbar joint centre was defined through a virtual marker created along the distance connecting the L5-S1 marker and the midpoint between the two ASIS markers (Seay et al., 2008), projected from the T12-L1 joint centre. The T12-L1 joint centre was defined using a virtual marker projected from the midpoint of the markers placed bilaterally on the ribcage at the T12-L1 joint space level onto the thorax longitudinal axis. The proximal end of this axis was defined as the midpoint between the suprasternal notch and the second thoracic vertebra, while the distal end was defined as the midpoint between the xiphoid process and the inferior angles of most caudal points of the two scapulae. At the pelvis, two markers were placed on each anterior and posterior superior iliac spine, as well as on top of each iliac crest. The proximal end was defined using the virtual marker created as the distal end of the lumbar segment and the distal end was defined as the midpoint between the hips. A second local coordinate system (LCS) (for kinematic computations only) was created based on the CODA pelvis model (Robertson et al., 2014) in order to achieve a more clinically recognisable pelvic tilt (sagittal plane). All the LCSs were defined in accordance with Robertson et al. (2014).

A Woltring generalized cross-validatory cubic spline smoothing routine (Woltring, 1986) was used to filter both kinematic and kinetic data. Lower limb and trunk joint angles (using a XYZ Cardan sequence) and moments (determined through inverse dynamics and normalized to subjects' body mass) were computed and expressed relatively to the proximal segment. Data was normalized to 100% stride cycle, time-distance parameters were normalized to subjects' height and joint moments to subjects' mass. Additionally peak values for hip and trunk joint angles and joint moments, as well as time-distance parameters, were computed for each cycle and averaged for each subject.

2.2.5 Data Analysis

The ICCs and their 95% confidence intervals for the 2-way mixed-effects model (Shrout & Fleiss, 1979) were calculated for anthropometric, time-distance, and key kinematic/kinetic parameters of left lower limb, which was chosen. A minimum of 0.80 was considered to be an acceptable ICC, i.e. 20% of the total variance was due to measurement error (or within subjects variance). Calculations also included the mean difference between measurements (D), and the 95% CI for D, the SD_{diff} and the 95% limits of agreement (95% LOA). The SEM and MDC were calculated using the following equations: $SEM = SD_{diff} / \sqrt{2}$ (de Vet, Terwee, Knol, & Bouter, 2006b) and $MDC_{95} = 1.96 \times \sqrt{2} \times SEM$ (de Vet et al., 2006a).

The ICC statistical analysis was conducted using SPSS (version 20.0; IBM, Chicago, IL) and a critical level of $p<.05$ was considered significant. Bland and Altman calculations and plots were performed using MedCalc Software bvba (version 13.3.3). The SEM and MDC were calculated using Microsoft Excel 2007(Microsoft Corp., Redmond, WA).

2.3 Results

The 14 participants (10 females and 4 males; 43.8±6.7 years; 68±14.0 kg; 166.9±29.0 cm) included in this study had a median score of 3.5 (1 to 9) and 3.0 (0 to 8) in NRS, respectively in the first and second assessment moments. They also had a median score of 21.5 (0 to 41) and 19.5 (2 to 41) in QBPDS-PT, in the same assessment moments, and their complaints lasted predominantly more than 24 months (11 participants). Patients had a low level of pain and disability and were clinically stable, as no significant differences were detected between assessment moments ($p>.05$).

The ICCs were ≥ 0.93 for anthropometric parameters (Table 1) except for lumbar (0.73, 95% CI 0.16 to 0.91) and right foot (0.87, 95% CI 0.58 to 0.96) segments' length.

Table 1 - Reliability values for anthropometric measurements in the pain group.

Anthropometric Parameter	ICC	(95% CI)	Mean	D	(95% CI)	SD_{diff}	95% LOA	SEM	MDC
Mass (kg)	1.00	0.996 to 1.00	68.14	0.03	-0.35 to 0.41	0.66	-1.26 to 1.32	0.47	1.29
Thoracic Segment Length (cm)	0.94	0.83 to 0.98	29.00	0.06	-0.35 to 0.47	0.71	-1.33 to 1.46	0.01	0.01
Lumbar Segment Length (cm)	0.73	0.16 to 0.91	8.27	-0.31	-0.98 to 0.36	1.16	-2.58 to 1.96	0.01	0.02
Pelvis Segment Depth (cm)	0.95	0.86 to 0.99	19.89	0.12	-0.26 to 0.50	0.66	-1.18 to 1.42	0.00	0.01
Inter ASIS Distance (cm)	0.97	0.91 to 0.99	28.40	-0.27	-0.88 to 0.35	1.06	-2.35 to 1.82	0.01	0.02
Right Tight Segment Length (cm)	0.93	0.78 to 0.98	39.92	0.00	-0.78 to 0.77	1.00	-2.63 to 2.62	0.01	0.03
Left Tight Segment Length (cm)	0.95	0.85 to 0.98	40.14	0.00	-0.93 to 0.30	1.00	-2.40 to 1.77	0.01	0.02
Right Leg Segment Length (cm)	0.95	0.85 to 0.98	38.61	0.00	-0.54 to 0.53	1.00	-1.83 to 1.81	0.01	0.02
Left Leg Segment Length (cm)	0.96	0.88 to 0.99	38.63	0.28	-0.19 to 0.76	0.82	-1.32 to 1.89	0.01	0.02
Right Foot Segment Length (cm)	0.86	0.58 to 0.96	12.81	0.07	-0.31 to 0.44	0.65	-1.21 to 1.34	0.00	0.01
Left Foot Segment Length (cm)	0.95	0.85 to 0.98	12.81	-0.09	-0.36 to 0.18	0.46	-0.99 to 0.81	0.00	0.01

ICC, intraclass correlation coefficient; 95% CI, 95% confidence interval for the ICC; mean, mean of measurements at time 1 and time 2; D, mean of the differences between measurements at time 1 and 2 and the 95% CI for D, the standard deviation of the differences (SD_{diff}); 95% LOA, Bland and Altman 95% limits of agreement; SEM, standard error of measurement; MDC, minimal detectable change.

For time-distance parameters, ICCs were also > 0.90 (Table 2), with the exception of left (0.86, 95% CI 0.55 to 0.95) and right (0.88, 95% CI 0.61 to 0.96) stance phases' duration. The SEM and MDC values were low.

Table 2 - Reliability values for time-distance parameters in the pain group.

Time-distance Parameter	ICC	(95% CI)	Mean	D	(95% CI)	SD_{diff}	95% LOA	SEM	MDC
Speed (m/s)	0.92	0.76 to 0.98	1.08	0.02	-0.03 to 0.06	0.08	-0.15 to 0.18	0.06	0.16
Cycle Time (s)	0.93	0.78 to 0.98	1.12	-0.01	-0.04 to 0.01	0.04	-0.10 to 0.07	0.03	0.09
Double Limb Support Time (s)	0.92	0.74 to 0.97	0.14	0.00	-0.01 to 0.01	0.02	-0.03 to 0.03	0.01	0.03
Stride Length (m)	0.93	0.79 to 0.98	1.20	0.00	-0.03 to 0.03	0.05	-0.09 to 0.09	0.03	0.09
Stride Width (m)	0.97	0.92 to 0.99	0.09	-0.01	-0.01 to 0.00	0.01	-0.02 to 0.01	0.01	0.02
Left Lower Limb									
Cycle Time (s)	0.93	0.78 to 0.98	1.12	-0.01	-0.04 to 0.01	0.04	-0.1 to 0.08	0.03	0.09
Stance Time (s)	0.86	0.55 to 0.95	0.69	0.00	-0.03 to 0.03	0.05	-0.09 to 0.09	0.03	0.09
Step Time (s)	0.92	0.76 to 0.97	0.56	0.00	-0.02 to 0.01	0.02	-0.05 to 0.04	0.02	0.04
Step Length (m)	0.94	0.81 to 0.98	0.60	0.00	-0.01 to 0.02	0.02	-0.04 to 0.05	0.02	0.05
Stride Length (m)	0.93	0.78 to 0.98	1.20	0.00	-0.03 to 0.02	0.05	-0.10 to 0.09	0.03	0.09
Right Lower Limb									
Cycle Time (s)	0.93	0.78 to 0.98	1.12	-0.01	-0.04 to 0.01	0.04	-0.10 to 0.07	0.03	0.09
Stance Time (s)	0.88	0.61 to 0.96	0.71	0.00	-0.03 to 0.02	0.04	-0.08 to 0.08	0.03	0.08
Step Time (s)	0.92	0.76 to 0.98	0.56	-0.01	-0.02 to 0.01	0.03	-0.06 to 0.04	0.02	0.05
Step Length (m)	0.92	0.75 to 0.97	0.60	0.00	-0.02 to 0.01	0.03	-0.05 to 0.05	0.02	0.05
Stride Length (m)	0.93	0.78 to 0.98	1.20	0.01	-0.02 to 0.04	0.05	-0.09 to 0.10	0.03	0.10

ICC, intraclass correlation coefficient; 95% CI, 95% confidence interval for the ICC; mean, mean of measurements at time 1 and time 2; D, mean of the differences between measurements at time 1 and 2 and the 95% CI for D, the standard deviation of the differences (SD_{diff}); 95% LOA, Bland and Altman 95% limits of agreement; SEM, standard error of measurement; MDC, minimal detectable change.

Most of joint angle peaks showed ICCs greater than 0.80, while the remaining showed ICCs between 0.70 and 0.80 (varying from 0.53 to 0.97), with the exception of lumbar left lateral bending (0.50, 95% CI -0.55 to 0.84) (Table 3). The SEM values were generally ≤ 2.5º and the MDC values ranged between 2.3º and 11.3º. Bland–Altman plots with 95% LOA are shown in Figure 2 and outliers are visible (e.g. thoracic flexion and lumbar right lateral bending). The mean of the differences between measurements at time 1 and 2 was ≤ 1º in all parameters, except for thoracic left/right rotation and knee internal/external rotation angles. Plots of kinematics waveforms in the sagittal, transverse and frontal plane are available (Figure 3).

Table 3 - Reliability values for kinematic parameters in the pain group.

Kinematic Parameter	ICC	(95% CI)	Mean	D	(95% CI)	SD_{diff}	95% LOA	SEM	MDC
Thoracic Joint Angle (°)									
Peak Flexion	0.91	0.73 to 0.97	-2.7	0,1	-0.85 to 0.98	1.6	-3.04 to 3.17	1.1	3.1
Peak Right Lateral Bending	0.89	0.65 to 0.96	3.5	-0,1	-0.87 to 0.68	1.3	-2.72 to 2.53	0.9	2.6
Peak Left Rotation	0.84	0.51 to 0.95	3.9	-1,5	-4.65 to 1.70	5.5	-12.3 to 9.30	3.9	10.8
Peak Extension	0.90	0.68 to 0.97	-5.3	-0,1	-1.04 to 0.93	1.7	-3.39 to 3.29	1.2	3.3
Peak Left Lateral Bending	0.80	0.36 to 0.93	-3.4	-0,3	-0.92 to 0.43	1.2	2.54 to 2.04	0.8	2.3
Peak Right Rotation	0.87	0.59 to 0.96	-5.1	-1,7	-4.52 to 1.05	4.8	-11.20 to 7.73	3.4	9.5
Lumbar Joint Angle (°)									
Peak Flexion	0.84	0.49 to 0.95	9.7	0.8	-1.98 to 3.51	4.8	-8.54 to 10.07	3.4	9.3
Peak Right Lateral Bending	0.82	0.44 to 0.94	1.9	-0.1	-1.14 to 0.85	1.7	-3.52 to 3.23	1.2	3.4
Peak Left Rotation	0.88	0.64 to 0.96	3.4	0.5	-2.00 to 3.02	4.4	-8.01 to 9.03	3.1	8.5
Peak Extension	0.88	0.62 to 0.96	7.0	0.6	-1.80 to 2.89	4.1	-7.42 to 8.52	2.9	7.9
Peak Left Lateral Bending	0.50	-0.55 to 0.84	-0.9	0.01	-1.25 to 1.26	2.2	-4.25 to 4.27	1.5	4.3
Peak Right Rotation	0.87	0.59 to 0.96	-0.8	0.3	-2.34 to 3.01	4.6	-8.74 to 9.41	3.3	9.1
Hip Joint Angle (°)									
Peak Flexion	0.86	0.57 to 0.96	26.1	0.6	-1.70 to 2.95	4.0	-7.25 to 8.50	2.8	7.9
Peak Abduction	0.71	0.10 to 0.91	3.7	1.00	-0.61 to 2.60	2.8	-4.45 to 6.45	1.9	5.5
Peak External Rotation	0.82	0.43 to 0.94	12.5	0.1	-3.21 to 3.42	5.7	-11.15 to 11.35	4.1	11.3
Peak Extension	0.90	0.67 to 0.97	-11.3	0.4	-1.67 to 2.53	3.6	-6.70 to 7.56	2.6	7.1
Peak Adduction	0.81	0.40 to 0.94	-9.9	-0.2	-1.32 to 0.97	1.9	-4.07 to 3.71	1.4	3.9
Peak Internal Rotation	0.84	0.50 to 0.95	-5.0	-0.4	-3.50 to 2.81	5.5	-11.06 to 10.37	3.9	10.7
Knee Joint Angle (°)									
Peak Flexion	0.82	0.45 to 0.94	61.4	0.3	-1.31 to 1.92	2.8	-5.18 to 5.78	1.9	5.5
Peak Abduction	0.91	0.71 to 0.97	6.6	-0.7	-2.21 to 0.87	2.7	-5.89 to 4.55	1.9	5.2
Peak External Rotation	0.84	0.49 to 0.95	7.5	2.2	0.39 to 3.99	3.1	-3.92 to 8.31	2.2	6.1
Peak Extension	0.73	0.15 to 0.91	-1.2	0.6	-1.50 to 2.64	3.6	-6.47 to 7.62	2.5	7.0
Peak Adduction	0.71	0.09 to 0.91	-3.2	-0.4	-2.56 to 1.68	3.7	-7.64 to 6.75	2.6	7.2
Peak Internal Rotation	0.88	0.62 to 0.96	-10.3	1.7	-0.49 to 3.84	3.8	-5.69 to 9.04	2.7	7.4

Table 3 (cont.) - Reliability values for kinematic parameters in the pain group.

Kinematic Parameter	ICC	(95% CI)	Mean	D	(95% CI)	SD_{diff}	95% LOA	SEM	MDC
Ankle Joint Angle (°)									
Peak Dorsiflexion	0.85	0.54 to 0.95	88.6	0.4	-0.98 to 1.76	2.4	-4.25 to 5.03	1.7	4.6
Peak Abduction	0.92	0.77 to 0.98	20.2	-0.7	-2.91 to 1.50	3.8	-8.18 to 6.78	2.7	7.5
Peak External Rotation	0.87	0.58 to 0.96	23.3	-0.2	-1.93 to 1.59	3.1	-6.15 to 5.82	2.2	5.9
Peak Plantar Flexion	0.97	0.90 to 0.99	61.4	0.5	-0.79 to 1.71	2.2	-3.78 to 4.70	1.5	4.2
Peak Adduction	0.87	0.61 to 0.96	3.3	-0.8	-2.86 to 1.23	3.5	-7.76 to 6.13	2.5	6.9
Peak Internal Rotation	0.83	0.47 to 0.95	12.9	-0.5	-2.07 to 1.09	2.7	-5.85 to 4.87	1.9	5.4

ICC, intraclass correlation coefficient; 95% CI, 95% confidence interval for the ICC; mean, mean of measurements at time 1 and time 2; D, mean of the differences between measurements at time 1 and 2 and the 95% CI for D, the standard deviation of the differences (SD_{diff}); 95% LOA, Bland and Altman 95% limits of agreement; SEM, standard error of measurement; MDC, minimal detectable change.

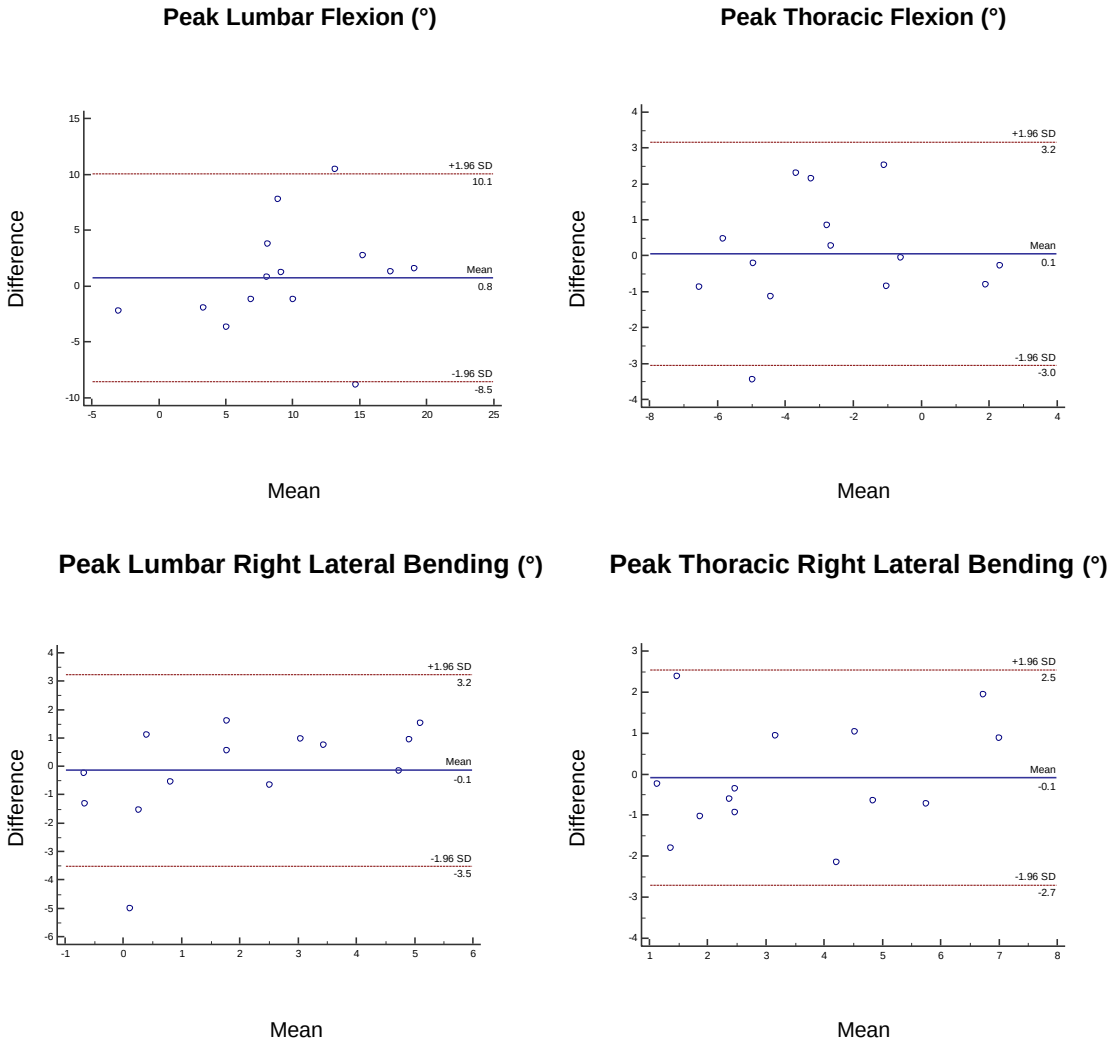


Figure 2 - Bland-Altman plots with 95% limits of agreement (dashed lines) for thoracic and lumbar peak joint angles in the pain group.

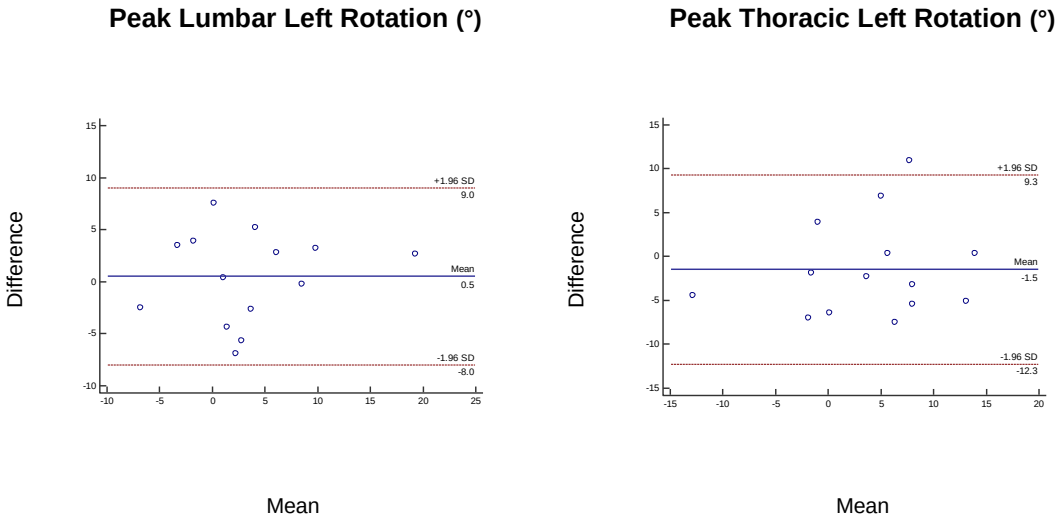


Figure 2 (cont.) - Bland-Altman plots with 95% limits of agreement (dashed lines) for thoracic and lumbar peak joint angles in the pain group.

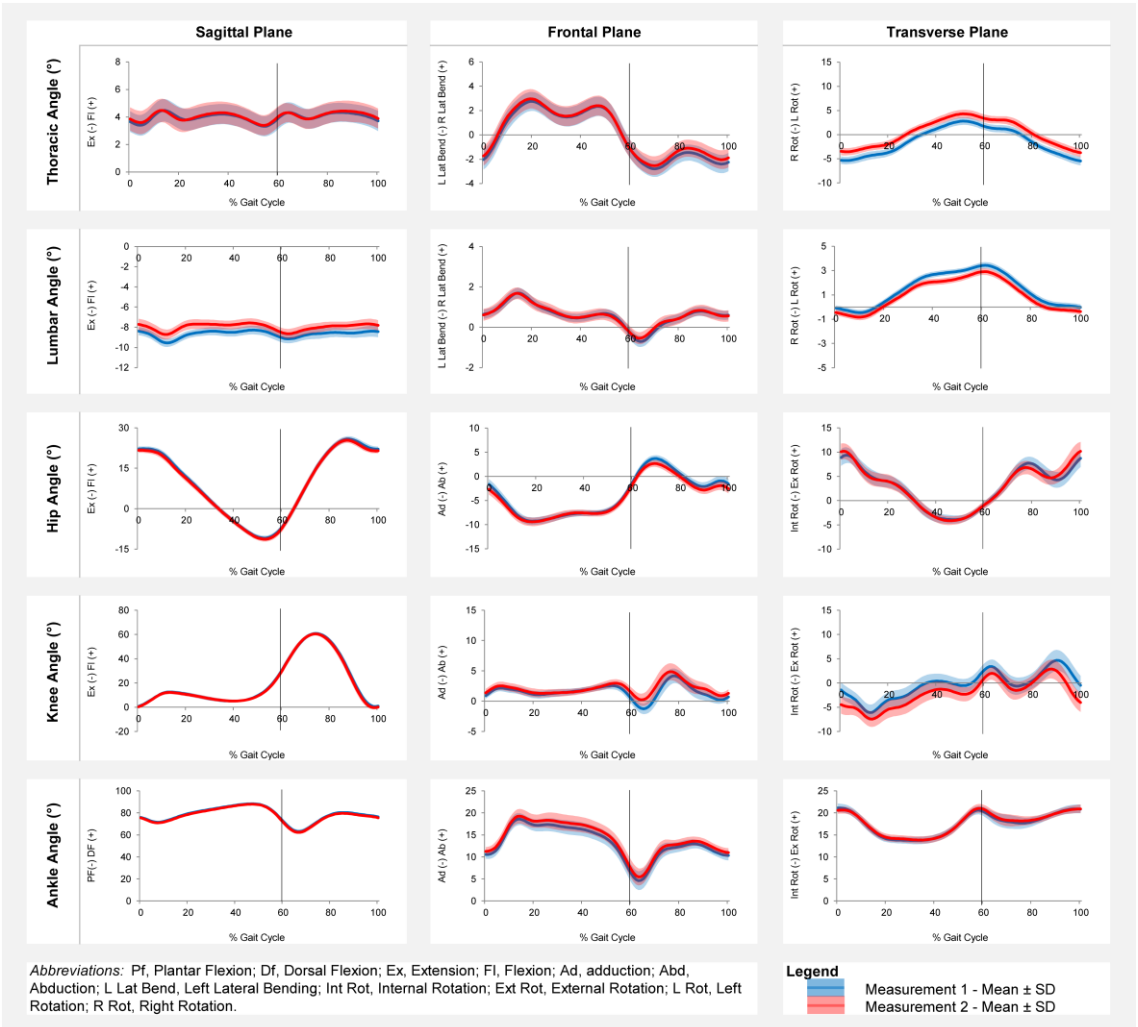


Figure 3 - Plots of joint angles waveforms during the gait cycle in the pain group.

For kinetic calculations one participant was excluded because of short stride length and failure in striking the force platform. In general, the ICC of kinetic parameters was lower than those for the kinematic data, but the majority was still above 0.7 (Table 4). Hip and ankle internal rotation moments showed the lowest values, 0.41 (CI 95% -0.94 to 0.82) and 0.38 (CI 95% -1.02 to 0.81), respectively. Generally SEM was ≤ 0.06 Nm/kg (varying from 0.01 to 0.12 Nm/kg) and MDC was ≤ 0.18 Nm/kg (varying from 0.06 to 0.33 Nm/kg).

Table 4 - Reliability values for kinetic parameters in the pain group.

Kinetic Parameter	ICC	(95% CI)	Mean	D	(95% CI)	SD_{diff}	95% LOA	SEM	MDC
Thoracic Joint Moment (Nm/kg)									
Peak Flexion	0.77	0.26 to 0.93	0.27	0.01	-0.06 to 0.09	0.12	-0.22 to 0.25	0.08	0.23
Peak Right Lateral Bending	0.68	-0.03 to 0.90	0.11	-0.04	-0.12 to 0.04	0.13	-0.31 to 0.22	0.09	0.26
Peak Left Rotation	0.85	0.51 to 0.95	0.08	0.02	-0.00 to 0.04	0.03	-0.05 to 0.08	0.02	0.06
Peak Extension	0.58	-0.38 to 0.87	-0.32	-0.01	-0.12 to 0.10	0.18	-0.36 to 0.34	0.12	0.33
Peak Left Lateral Bending	0.70	0.03 to 0.91	-0.31	-0.02	-0.11 to 0.07	0.15	-0.30 to 0.27	0.10	0.28
Peak Right Rotation	0.75	0.19 to 0.92	-0.12	-0.02	-0.04 to 0.00	0.04	-0.09 to 0.05	0.03	0.07
Lumbar Joint Moment (Nm/kg)									
Peak Flexion	0.87	0.58 to 0.96	0.21	0.01	-0.05 to 0.07	0.10	-0.18 to 0.20	0.07	0.18
Peak Right Lateral Bending	0.73	0.11 to 0.92	0.12	-0.05	-0.13 to 0.03	0.14	-0.32 to 0.22	0.09	0.26
Peak Left Rotation	0.79	0.32 to 0.94	0.09	0.01	-0.01 to 0.04	0.03	-0.05 to 0.08	0.02	0.06
Peak Extension	0.71	0.04 to 0.91	-0.36	-0.01	-0.10 to 0.09	0.16	-0.31 to 0.30	0.11	0.30
Peak Left Lateral Bending	0.65	-0.14 to 0.89	-0.33	-0.03	-0.13 to 0.07	0.16	-0.35 to 0.29	0.11	0.31
Peak Right Rotation	0.76	0.21 to 0.93	-0.13	-0.01	-0.03 to 0.02	0.04	-0.09 to 0.08	0.03	0.07
Hip Joint Moment (Nm/kg)									
Peak Flexion	0.90	0.67 - 0.97	0.51	0.05	0.00 to 0.10	0.08	-0.11 to 0.21	0.06	0.16
Peak Abduction	0.91	0.70 - 0.97	0.93	0.06	0.01 to 0.11	0.09	-0.11 to 0.23	0.06	0.17
Peak External Rotation	0.84	0.48 - 0.95	0.13	0.03	0.00 to 0.06	0.04	-0.05 to 0.11	0.03	0.08
Peak Extension	0.88	0.59 - 0.96	-0.41	-0.01	-0.06 to 0.04	0.07	-0.16 to 0.14	0.05	0.14
Peak Adduction	0.70	0.03 - 0.91	-0.11	0.00	-0.02 to 0.02	0.03	-0.06 to 0.06	0.02	0.06
Peak Internal Rotation	0.41	-0.94 - 0.82	-0.09	-0.01	-0.05 to 0.02	0.05	-0.12 to 0.09	0.04	0.10
Knee Joint Moment (Nm/kg)									
Peak Flexion	0.75	0.16 - 0.92	0.33	-0.01	-0.06 to 0.03	0.07	-0.16 to 0.13	0.05	0.14
Peak Abduction	0.65	-0.15 - 0.89	0.35	0.04	-0.01 to 0.09	0.08	-0.12 to 0.21	0.06	0.16
Peak External Rotation	0.87	0.58 - 0.96	0.10	0.00	-0.01 to 0.02	0.03	-0.05 to 0.05	0.02	0.05
Peak Extension	0.88	0.61 - 0.96	-0.39	-0.03	-0.08 to 0.02	0.08	-0.19 to 0.13	0.06	0.16
Peak Adduction	0.89	0.64 - 0.97	-0.04	-0.01	-0.02 to 0.00	0.02	-0.04 to 0.02	0.01	0.03
Peak Internal Rotation	0.52	-0.57 - 0.85	-0.10	-0.01	-0.04 to 0.01	0.04	-0.09 to 0.07	0.03	0.08

Table 4 (cont.) - Reliability values for kinetic parameters in the pain group.

Kinetic Parameter	ICC	(95% CI)	Mean	D	(95% CI)	SD_{diff}	95% LOA	SEM	MDC
Ankle Joint Moment (Nm/kg)									
Peak Dorsiflexion	0.68	-0.06 - 0.90	0.07	0.00	-0.02 to 0.02	0.03	-0.07 to 0.06	0.02	0.06
Peak Abduction	0.61	-0.29 - 0.88	0.04	0.01	-0.01 to 0.03	0.03	-0.04 to 0.07	0.02	0.05
Peak External Rotation	0.94	0.79 - 0.98	0.13	0.01	0.00 to 0.03	0.02	-0.03 to 0.06	0.02	0.04
Peak Plantar Flexion	0.88	0.59 - 0.96	-1.37	0.00	-0.03 to 0.04	0.06	-0.11 to 0.11	0.04	0.11
Peak Adduction	0.91	0.70 - 0.97	-0.35	0.05	-0.02 to 0.11	0.11	-0.16 to 0.25	0.07	0.20
Peak Internal Rotation	0.38	-1.02 - 0.81	-0.03	0.01	-0.01 to 0.02	0.02	-0.03 to 0.04	0.01	0.03

ICC, intraclass correlation coefficient; 95% CI, 95% confidence interval for the ICC; mean, mean of measurements at time 1 and time 2; D, mean of the differences between measurements at time 1 and 2 and the 95% CI for D, the standard deviation of the differences (SD_{diff}); 95% LOA, Bland and Altman 95% limits of agreement; SEM, standard error of measurement; MDC, minimal detectable change.

2.4 Discussion

To the best of our knowledge this is the first study that evaluated test-retest reliability and MDC of 3DGA in CLBP patients. The results show high test-retest reliability for lower limb and trunk kinematics and time distance parameters during gait in CLBP individuals, together with a clinically acceptable level of error.

Concerning the 10-cycle average for joint angle parameters, most of the parameters showed ICC values greater than 0.80, which means that just 20% of the obtained variance was due to either measurement error or within-subject variability over time. Some authors recommend a minimum ICC value of 0.70 for research purposes (Nunnaly, 1978) while others defend that this value cannot be set in absolute terms and that should be taken into account the aim of the instrument under investigation (Streiner & Norman, 2008). In any case, such high ICC values demonstrate high reliability for kinematics data during 3DGA in CLBP, including for the lumbar and thoracic segments. The mean of the differences between the two assessments was small and no systematic bias was detected (except for knee external rotation), as zero was included in the 95% CI. Joint angles also did not show indications of heteroscedasticity, i.e., larger variability for higher test values, which reinforces the agreement of the measurements. Standard error of measurement was small (≤ 2.5º) for the majority of the parameters and MDC values were generally higher in transverse plane parameters. The 95% LOA intervals were relatively wide, pointing out that a substantial difference in an individual joint angle would be required to allow us to confidently state that a real individual change had taken place (Mieritz et al., 2014). Therefore, these measurements may be particularly useful and appropriate for detecting change in groups of subjects.

The absence of data on test-retest reliability and MDC of 3DGA in CLBP patients precludes comparisons with standard data regarding this population. Nevertheless, a systematic review (McGinley et al., 2009) examining the reliability of three-dimensional kinematic gait measurements in healthy individuals and in individuals with pathology, such as stroke or cerebral palsy, reported error values between 2º and 5º. With the exception of the hip, lumbar and thoracic transverse plane parameters, the error values of our study fell between 1º and 3º, which is in line with the mentioned study. Accordingly to McGinley et al. (2009), errors between 2º and 5º are likely to be regarded as reasonable although may require consideration in data interpretation, which suggests that all of our kinematic parameters results have an acceptable clinical level of error and its use can be considered. Regarding CLBP patients' movement strategies, reliability of thoracic, lumbar and hip kinematic parameters might be particularly important, however, care should be taken when interpreting transverse plane parameters.

Reliability of simple trunk motions with 3D regional spinal motion instruments in CLBP patients has also been evaluated (Mieritz et al., 2012). These authors conducted a systematic review and found ICC based reliability coefficients above 0.7 for most of the reviewed motion parameters. However, due to the lack of information on methodological issues, they reported that reliability estimate is difficult to interpret. A recent study with a large sample size (220 subjects), reported that kinematic data on the spinal movement in the sagittal plane may be sufficiently reliable in measurements of groups of CLBP patients (Mieritz et al., 2014).

The reliability of the average value for joint moments was lower than for joint angles, but most parameters still showed an ICC > 0.7 (varying from 0.38 to 0.94). Correlation coefficients might be affected by the range of variation of the parameter within the sample and large variations between subjects can result in higher ICC values (Rankin & Stokes, 1998). The SDs of the mean values of joint moments parameters were smaller than those of joint angles, which may have contributed to their reduced variation and low ICC values. Despite the lower ICC values, joint moments' SEM and MDC were still low. No systematic bias or heteroscedasticity was detected, which emphasize the agreement regarding joint moments parameters.

Anthropometric measurements showed high ICC values (≥ 0.93), except for lumbar (0.73, 95% CI 0.16 to 0.91) and right foot (0.87, 95% CI 0.58 to 0.96) segments' length, which may be representative of high reliability of marker placement, a major source of

variability in test-retest experimental procedures of 3DGA studies (McGinley et al., 2009). Another potential source of variability in test-retest gait analysis experiments is the relative skin marker movement error that has been shown to affect the accuracy of calculated joint kinematics, especially in the frontal and transverse planes (Cappozzo, Catani, Leardini, Benedetti, & Croce, 1996). Different soft tissue artifact compensation methods, such as optimization techniques, have been proposed. In this study we used the segment optimization (SO) method that estimates each segment pose independently by finding the optimal fit, in a least-squares sense, between the model determined and the measured markers coordinates (Challis, 1995). This method has proven to be more effective for computing the position and orientation of body segments comparatively to the direct method (Robertson et al., 2014), which may have contributed to high reliability and low level or error of the measurements.

Time-distance parameters also showed excellent reliability and agreement, which is a good indicator of true stability between measurement sessions (McGinley et al., 2009). Additionally, 95% LOA intervals were very restricted, indicating that these parameters are appropriate for detecting individual and group changes, which is important both for clinical and research contexts.

Strengths and limitations:

The strengths of the current study is that 2 measurement times were separated by 7 days, with all evaluations occurring at the same time of the day to diminish interference of recall bias. Since the assessor was blinded to the results of the first assessment, its interference was limited. Testing conditions also contributed to high reliability of our measurements, with participants allowed to adopt their normal gait (Monaghan et al., 2007) and very stable pace. Since reliability may vary across differing levels of LBP severity (Streiner & Norman, 2008), we included participants with different levels of pain intensity (NRS score 0 to 8) and disability (QBPDS-PT score 0 to 41). Therefore, we are confident to state that our data offer a robust estimate of the reliability and measurement error of lower limb and trunk data during gait in CLBP. There are a few limitations in this study. We chose a convenience sample from a prospective study that agreed to perform two consecutive evaluations. The small sample led to wide 95% CIs for ICCs, which may contribute to some uncertainty in the findings. There is also lack of information regarding inter-tester reliability, so care should be taken when extending our results to assessments conducted by several testers or in other laboratories.

Future studies should focus on the responsiveness of 3DGA, in order to evaluate the capability of this method in detecting change in a patient's condition as a result of an intervention or to distinguish individual differences in response to treatment.

2.5 Conclusions

The results of this study show high test-retest reliability for lower limb and trunk joint angles and time-distance parameters during gait in CLBP individuals, together with a low measurement error. These results also support the use of this method in clinical assessments of CLBP patients' gait patterns.

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3

Three-dimensional multi-segmental trunk kinematics and kinetics during gait: Test-retest reliability and minimal detectable change

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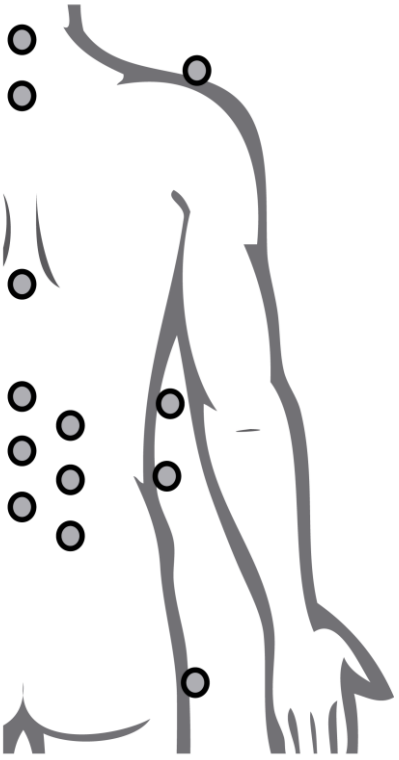

Abstract

Background and Aim: Trunk kinematics and kinetics can contribute to more detailed information on gait impairment, however, data about reliability and measurement error of multi-segment trunk on 3DGA is lacking. The aim of this study is to investigate test-retest reliability and MDC of 3DGA kinematic and kinetic data in a sample of healthy individuals, using a two rigid segment model for the trunk.

Methods A test-retest study with a median interval of 7 days and a sample of 23 healthy individuals was conducted. Anthropometric, time-distance parameters and peak values for lower limb and trunk joint angles/moments were computed. The $ICC_{3,k}$, SEM, MDC and 95% LOA were calculated.

Results: Acceptable test-retest reliability for most joint angles and a SEM ≤4°. The ICCs were above 0.7 for joint moments and the SEM and MDC were ≤0.2 Nm/kg and ≤0.6 Nm/kg, respectively. Bland-Altman plots with 95% LOA revealed a good agreement and time-distance parameters were all highly repeatable (majority ICCs > 0.90).

Conclusions: The results of this study suggest varied reliability indices for multi-segment trunk joint angles and moments during gait and an acceptable level of error, particularly for sagittal plane parameters. Some parameters showed wide 95% CIs for ICCs and higher SEM%. However, we believe this study provide preliminary data regarding reliability indices for multi-segment trunk during gait, which may be valuable for clinical reasoning and decision making when dealing with musculoskeletal disorders.

Keywords

Three Dimensional Gait Analysis; Multi-segmental trunk; Reliability; Measurement Error.

3.1 Introduction

Three-dimensional gait analysis is a valuable assessment method used in clinical and in research settings to support clinical functional diagnoses and decision-making. Repeated gait measurements can also be useful to evaluate the outcome of therapeutic interventions, although the observed variability between pre and post intervention measurements may be due to treatment effects or measurement variation, or a combination of both (McGinley, Baker, Wolfe, & Morris, 2009). Thus, knowledge about the error magnitude can minimise the risk of over-interpreting small differences as meaningful (Schwartz, Trost, & Wervey, 2004) and can contribute to the certainty that a measured intervention effect exceeds the measurement error. In 3DGA there are numerous potential sources of variability affecting the error magnitude of the testing procedure, such as instrumental errors, anatomical landmark misplacement and STA (Cereatti, Della Croce, & Cappozzo, 2006).

Knowledge about reliability and MDC values from healthy population is extremely important since it can help clinicians and researchers interpreting pathological data. Several studies have investigated the reliability of 3DGA in healthy and pathological populations (McGinley et al., 2009). A systematic review examining the reliability of three-dimensional kinematic gait measurements in healthy individuals and in individuals with pathology (such as stroke or cerebral palsy) reported a variable median value of within-assessor reliability (0.54 to 0.96) (McGinley et al., 2009). Additionally, they revealed error values between 2º and 5º, concluding that although most errors in gait analysis are probably acceptable, they are generally not small enough to be ignored during clinical data interpretation.

Despite the importance of such information, only two studies (Meldrum, Shouldice, Conroy, Jones, & Forward, 2014; Wilken, Rodriguez, Brawner, & Darter, 2012) provided absolute measures of measurement error and MDC values for kinematic and kinetic parameters in healthy individuals. Meldrum et al. (2014) reported low SEM (≤5˚) for the majority of the lower limb kinematic parameters and variable ICCs values (0.14 to 0.92). They also described the repeatability of key kinetic gait cycle parameters, predominantly in the sagittal plane (except for hip abductor joint moment), showing ICCs that varied from 0.51 to 0.81. Using a sample of young healthy adults, Wilken et al. (2012) reported good to excellent reliability of lower limb and trunk kinematics/kinetics across a range of controlled walking velocities, as well as low MDC values (approximately of 5º for joint angles). By adding trunk data, this study made an important contribution, given the established relevance of the coordination between

trunk and pelvis rotations, as well as trunk muscle activity during normal walking (Lamoth, Meijer, Daffertshofer, Wuisman, & Beek, 2006). It contributes for the maintenance of dynamic equilibrium, reduces the energy cost and helps to effectively deal with perturbations during locomotion (Lamoth et al., 2006). However, in Wilken´s study, the trunk was modelled as one rigid segment and kinetic transverse plane parameters were lacking, which excludes valuable information for clinical reasoning and decision making when dealing with musculoskeletal disorders. Thus, the aim of this study is to investigate test-retest reliability and MDC of 3DGA kinematic and kinetic data in a sample of healthy individuals, using a two rigid segment model for the trunk.

3.2 Materials and Methods

3.2.1 Study Design

A prospective within assessor test-retest study was conducted.

3.2.2 Sample Size Calculation

The sample size calculation for a pre-defined 5% level of significance with 80% power was performed using the formula of Kraemer and Thiemann (Kraemer & Thiemann, 1987). The desired reliability co-efficient was set at 0.90 with a minimum reliability of 0.70. This resulted in a sample size requirement of 17, however, to allow for non-attenders and increased precision 23 subjects were recruited.

3.2.3 Participants

A convenience sample of 23 volunteers (12 females and 11 males; age 35±7.3 years, height 1.70±0.07 m, mass 66.39±9.2 kg and body mass index 23.01± 2.3 kg/m^2) was recruited from university staff and their associates to participate in a 12-week prospective study, according to a standardized recruitment protocol. Firstly, physiotherapists from the research team carried out individuals' recruitment based on predefined inclusion/exclusion criteria. Healthy individuals were considered eligible if they were aged between 18 and 65 years old and were excluded if they had any clinical condition (musculoskeletal, neurological, cardiac or pulmonary) or symptom that could affect gait. Pregnant women were also excluded. After this screening, from the initial pool of 35 participants, 23 were recruited (one was excluded because of depression diagnosis and 11 were not available to perform two consecutive assessments with a mean interval of 7 days due to the lack of time).

The local Ethics Committee approved the study. All the participants were informed of the procedures and risks of the study and signed an informed consent.

3.2.4 Procedures

Gait analysis was performed twice with an interval of 7 to 11 days (median of 7 days). This time interval was considered long enough to avoid assessor memory bias and short enough to avoid a change in individuals' gait pattern (McDermott, Bolger, Keating, McEvoy, & Meldrum, 2010). On the first visit to the laboratory, participants' history was reviewed. This was complemented with the measurement of body mass and height. Body segments' length was obtained using the respective proximal and distal anatomical landmarks collected during the static trial described below. For pelvis, ASIS and PSIS markers were used.

Finally, gait data was collected using a 13-camera opto-electronic system (Oqus 300, Qualisys AB, Gothenburg, Sweden) synchronized in time and space with two Kistler (9281B and 9283U014, Kistler Group, Winterthur, Switzerland) and one AMTI (BP6001200, Advanced Mechanical Technology, Inc Watertown, USA) force platforms, at 200 Hz. The marker set used was based on previous reports (Leardini, Biagi, Merlo, Belvedere, & Benedetti, 2011; Seay, Selbie, & Hamill, 2008) (Figure 1). After a static trial, participants were instructed to walk barefoot at their preferred walking speed, continuously and during short periods of time (1-2 minutes) to avoid fatigue. A familiarization trial was performed before data collection. Each participant was assessed at the same time of the day to minimize the effects of diurnal variations in joint mechanics. All the procedures were carried out by the same assessor.

3.2.4 Data Processing

Considering the natural variability in kinematic and kinetic gait parameters, 10 cycles were selected (Monaghan, Delahunt, & Caulfield, 2007). Cycles were extracted using Qualysis Track Manager (v2.8 build 1554, Qualisys AB, Gothenburg, Sweden) and exported to be processed under Visual 3D software (v5.01.10, C-Motion, Inc, Rockville, USA).

A 9-segment model (feet, shanks, thighs, pelvis, lumbar and thoracic spine) was built for each participant (Leardini et al., 2011; Seay et al., 2008). All the LCSs were defined in accordance with Robertson, Caldwell, Hamill, Kamen, & Whittlesey (2014) and their origin was the joint centre. The ankle and knee joint centres were defined as the midpoint of the tibia malleoli and as the midpoint of the femur epicondyles, respectively (Robertson et al., 2014). The hip joint centres were computed using the pelvis markers,

according to published regression equations (Bell, Pedersen, & Brand, 1990). The lumbar joint centre was defined through a virtual marker created along the distance connecting the L5-S1 marker and the midpoint between the two ASIS markers (Seay et al., 2008), projected from the thoracic joint centre. The thoracic joint centre was defined using a virtual marker projected from the midpoint of the markers placed bilaterally on the ribcage at the T12-L1 joint space level onto the thorax longitudinal axis. The proximal end of this axis was defined as the midpoint between the suprasternal notch and the second thoracic vertebra, while the distal end was defined as the midpoint between the xiphoid process and the inferior angles of most caudal points of the two scapulae. At the pelvis, two markers were placed on each anterior and posterior superior iliac spine, as well as on top of each iliac crest. The proximal end was defined using the virtual marker created as the distal end of the lumbar segment and the distal end was defined as the midpoint between the hips. A second LCS (for kinematic computations only) was created based on the CODA pelvis model (Robertson et al., 2014) in order to achieve a more clinically recognisable pelvic tilt (sagittal plane). Each segment was considered to be independent and to have 6 degrees of freedom (SO method) (Cappello, La Palombara, & Leardini, 1996). Lower limb segment masses were determined according to Dempster (1955) while the remaining inertial parameters were computed based on Hanavan (1964). Lumbar and thoracic inertial parameters were computed according to Pearsall, Reid and Livingston (1996).

A Woltring generalized cross-validatory cubic spline smoothing routine (Woltring, 1986) with an error variance of 0.0001 (mean standard error of 1 mm) was used to filter kinematic and kinetic data. Lower limb and trunk joint angles (using a XYZ Cardan sequence) and joint moments (determined through inverse dynamics and normalized to subjects' body mass) were computed consistent with Robertson et al. (2014; pp 50-54 and 152-164) and expressed relatively to the proximal segment. Thus, flexion/extension rotations occurred around the medio-lateral axis of the proximal segment, abduction/adduction/lateral bending rotations around a floating axis and external/internal/right/left rotations around the distal segment longitudinal axis. Data was normalized to 100% stride cycle, time-distance parameters were normalized to subjects' height and joint moments to subjects' mass. Additionally peak values for hip and trunk joint angles and joint moments, as well as time-distance parameters, were computed for each cycle and averaged for each subject.

3.2.5 Data Analysis

The ICC and their 95% confidence intervals for the 2-way mixed-effects model (Shrout & Fleiss, 1979) were calculated for anthropometric, time-distance, and key kinematic and kinetic parameters. Left lower limb segments were chosen for data analysis, as it was not expected any systematic difference in reliability between lower limbs in healthy individuals (Meldrum et al., 2014). A minimum of 0.70 was considered to be an acceptable ICC. Calculations also included the mean difference between measurements (D), and the 95% CI for D, the SD_{diff} and the 95% limits of agreement (95% LOA). The SEM and MDC were calculated using the following equations: SEM = $SD_{diff}/\sqrt{2}$ (de Vet, Terwee, Knol, & Bouter, 2006b) and MDC_{95} = 1.96 x $\sqrt{2}$ x SEM (de Vet et al., 2006a).

The ICC statistical analysis was conducted using SPSS (version 20.0; IBM, Chicago, IL) and a critical level of p <. 05 was considered significant. Bland and Altman calculations were performed using MedCalc Software bvba (version 13.3.3). The SEM, MDC and joint angles/moments waveforms were computed using Microsoft Excel 2007 (Microsoft Corp., Redmond, WA).

3.3 Results

For anthropometric parameters, the ICCs between the first and second measurement were ≥0.80 (Table 5), with the exception of lumbar length (0.79, 95% CI 0.5 to 0.91). Time-distance parameters also showed ICCs between measurements above the minimum acceptable level and many were even above 0.90 (Table 5).

Table 5 - Reliability values for anthropometric and time-distance measurements in the control group.

Anthropometric Parameter	ICC	(95% CI)	Mean	Mean (Min to Max)	D	(95% CI)	SD_{diff}	95% LOA	SEM	MDC
Mass (kg)	1.00	1 to 1	66.42	51.5 to 82.2	-0.28	-0.52 to 0.05	0.55	-1.36 to 0.79	0.39	1.07
Height (m)	1.00	1 to 1	1.70	1.5 to 1.8	0.00	0 to 0	0.00	-0.01 to 0	0.00	0.01
Thoracic Segment Length (cm)	0.89	0.74 to 0.95	0.30	0.3 to 0.3	0.01	0 to 0.01	0.01	-0.01 to 0.03	0.01	0.02
Lumbar Segment Length (cm)	0.79	0.5 to 0.91	0.09	0.1 to 0.1	0.00	-0.01 to 0.01	0.01	-0.03 to 0.03	0.01	0.03
Pelvis Segment Depth (cm)	0.83	0.59 to 0.93	0.19	0.2 to 0.2	0.00	0 to 0	0.01	-0.01 to 0.02	0.01	0.02
Inter ASIS Distance (cm)	0.90	0.76 to 0.96	0.26	0.2 to 0.3	0.00	-0.01 to 0.01	0.01	-0.03 to 0.03	0.01	0.03
Left Tight Segment Length (cm)	0.96	0.9 to 0.98	0.41	0.4 to 0.4	0.00	-0.01 to 0	0.01	-0.02 to 0.01	0.01	0.02
Left Shank Segment Length (cm)	0.94	0.85 to 0.97	0.40	0.4 to 0.4	0.00	0 to 0.01	0.01	-0.02 to 0.02	0.01	0.02
Left Foot Segment Length (cm)	0.89	0.73 to 0.95	0.13	0.1 to 0.1	0.00	0 to 0	0.01	-0.01 to 0.01	0.00	0.01

Table 5 (cont.) - Reliability values for anthropometric and time-distance measurements in the control group.

Anthropometric Parameter	ICC	(95% CI)	Mean	Mean (Min to Max)	D	(95% CI)	SD_{diff}	95% LOA	SEM	MDC
Time-distance Parameter										
Speed (m/s)	0.94	0.85 to 0.97	1.20	1 to 1.4	0.01	-0.01 to 0.04	0.06	-0.11 to 0.13	0.04	0.12
Cycle Time (s)	0.94	0.85 to 0.97	1.08	1 to 1.2	-0.01	-0.03 to 0.01	0.04	-0.09 to 0.07	0.03	0.08
Double Limb Support Time (s)	0.89	0.75 to 0.95	0.11	0.1 to 0.1	0.00	-0.01 to 0	0.01	-0.02 to 0.02	0.01	0.02
Stride Length (m)	0.92	0.8 to 0.96	1.28	1.1 to 1.4	0.01	-0.01 to 0.02	0.04	-0.08 to 0.09	0.03	0.08
Stride Width (m)	0.90	0.76 to 0.96	0.09	0.1 to 0.1	0.00	0 to 0	0.01	-0.02 to 0.02	0.01	0.02
Left Lower Limb										
Cycle Time (s)	0.94	0.85 to 0.97	0.64	1 to 1.3	-0.01	-0.03 to 0.01	0.04	-0.09 to 0.07	0.03	0.08
Stance Time (s)	0.88	0.71 to 0.95	0.64	0.5 to 0.8	-0.01	-0.03 to 0	0.04	-0.09 to 0.06	0.03	0.07
Step Time (s)	0.90	0.78 to 0.96	0.54	0.5 to 0.6	0.00	-0.01 to 0.01	0.02	-0.05 to 0.05	0.02	0.05
Step Length (m)	0.82	0.57 to 0.92	0.65	0.6 to 0.7	0.01	-0.01 to 0.02	0.03	-0.06 to 0.07	0.02	0.06
Stride Length (m)	0.92	0.81 to 0.97	1.28	1.1 to 1.4	0.01	-0.01 to 0.02	0.04	-0.08 to 0.09	0.03	0.08

ICC, intraclass correlation coefficient; 95% CI, 95% confidence interval for the ICC; Mean, mean of measurements at time 1 and time 2; Mean (Min to Max), minimum and maximum mean value; D, mean of the differences between measurements at time 1 and 2 and the 95% CI for D; the standard deviation of the differences (SD_{diff}); 95% LOA, Bland and Altman 95% limits of agreement; SEM, standard error of measurement; MDC, minimal detectable change.

Reliability of key kinematic parameters was examined and almost half of joint angle peaks had ICCs ≥0.80 (varying between 0.80 and 0.95) (Table 6). The ICC of the remaining parameters ranged between 0.60 and 0.80, with the exception of thoracic left rotation (0.51, 95% CI - 0.15 to 0.79). The lumbar segment reliability was higher for the sagittal plane, showing ICCs of 0.91 (95% CI 0.79 to 0.96) and 0.90 (95% CI, 0.77 to 0.96) for flexion and extension, respectively. Reliability of thoracic kinematics was slightly poorer for transverse planes, with ICCs varying between 0.51 and 0.62. The SEM was ≤4º for all parameters, except for hip rotations (varying from 5º to 6º). The MDC ranged from 2º to 16º and transverse plane parameters showed the greatest values, particularly internal (15º) and external (16º) hip rotations. The mean of the differences between measurements times was ≤1º in all parameters (Table 6).

Table 6 - Reliability values for kinematic parameters in the control group.

Kinematic Parameter	ICC	(95% CI)	Mean	Mean (Min to Max)	D	(95% CI)	SD_{diff}	95% LOA	SEM	MDC
Thoracic Joint Angle (°)										
Peak Flexion	0.71	0.32 to 0.88	4.9	0.2 to 10.2	0.39	-0.85 to 1.63	2.9	-5.22 to 6	2.0	5.6
Peak Right Lateral Bending	0.90	0.77 to 0.96	2.9	0.7 to 8.8	0.48	0 to 0.97	1.1	-1.72 to 2.68	0.8	2.2
Peak Left Rotation	0.51	-0.15 to 0.79	7.0	-0.4 to 16.9	0.49	-1.85 to 2.84	5.4	-10.15 to 11.14	3.8	10.6
Peak Extension	0.66	0.2 to 0.86	2.2	-1.8 to 7.1	0.40	-0.89 to 1.69	3.0	-5.44 to 6.24	2.1	5.8
Peak Left Lateral Bending	0.77	0.45 to 0.9	-3.5	-7.8 to -0.1	0.40	-0.22 to 1.02	1.4	-2.39 to 3.19	1.0	2.8
Peak Right Rotation	0.62	0.1 to 0.84	-3.8	-13.1 to 3.1	0.31	-1.9 to 2.51	5.1	-9.69 to 10.31	3.8	10.6

Table 6 (cont.) - Reliability values for kinematic parameters in the control group.

Kinematic Parameter	ICC	(95% CI)	Mean	Mean (Min to Max)	D	(95% CI)	SD_{diff}	95% LOA	SEM	MDC
Lumbar Joint Angle (°)										
Peak Flexion	0.91	0.79 to 0.96	-6.5	-19.1 to 7.6	0.57	-1.14 to 2.28	4.0	-7.17 to 8.31	2.8	7.7
Peak Right Lateral Bending	0.63	0.13 to 0.84	1.9	-1.1 to 5.6	-0.85	-1.86 to 0.16	2.3	-5.43 to 3.73	1.7	4.6
Peak Left Rotation	0.67	0.22 to 0.86	0.9	-7.2 to 7.8	-0.69	-2.69 to 1.32	4.6	-9.79 to 8.41	3.3	9.1
Peak Extension	0.90	0.77 to 0.96	-9.0	-21.9 to 5.6	0.92	-0.92 to 2.77	4.3	-7.45 to 9.3	3.0	8.4
Peak Left Lateral Bending	0.61	0.07 to 0.83	-2.0	-5.2 to 0.8	-0.87	-1.8 to 0.05	2.1	-5.06 to 3.31	1.5	4.2
Peak Right Rotation	0.64	0.15 to 0.85	-4.3	-13.4 to 2	-0.52	-2.63 to 1.58	4.9	-10.06 to 9.02	3.4	9.5
Hip Joint Angle (°)										
Peak Flexion	0.82	0.57 to 0.92	27.0	13.1 to 38.7	-0.34	-2.24 to 1.57	4.4	-8.97 to 8.3	3.1	8.6
Peak Abduction	0.68	0.25 to 0.87	7.5	3.4 to 11.3	0.83	-0.19 to 1.84	2.4	-3.78 to 5.44	1.7	4.6
Peak External Rotation	0.64	0.15 to 0.85	16.0	7.2 to 31.7	-0.62	-3.92 to 2.67	7.6	-15.55 to 14.3	5.4	14.9
Peak Extension	0.81	0.55 to 0.92	-12.2	-20 to -4.3	-0.24	-2.01 to 1.52	4.1	-8.23 to 7.74	2.9	8.0
Peak Adduction	0.77	0.46 to 0.9	-7.1	-13.3 to -1.8	0.44	-0.77 to 1.65	2.8	-5.05 to 5.93	2.0	5.5
Peak Internal Rotation	0.75	0.41 to 0.89	-1.7	-15 to 11.5	-1.09	-4.68 to 2.51	8.3	-17.37 to 15.2	5.9	16.3
Knee Joint Angle (°)										
Peak Flexion	0.87	0.7 to 0.95	62.3	58.6 to 68.4	0.50	-0.34 to 1.35	2.0	-3.33 to 4.34	1.4	3.8
Peak Abduction	0.75	0.42 to 0.9	9.1	0.1 to 17.9	-0.29	-2.39 to 1.81	4.9	-9.8 to 9.22	3.4	9.5
Peak External Rotation	0.82	0.58 to 0.93	2.8	-7.3 to 13.2	-0.11	-2 to 1.78	4.4	-8.68 to 8.45	3.1	8.6
Peak Extension	0.69	0.27 to 0.87	-1.5	-6.2 to 2.2	0.08	-1.06 to 1.21	2.6	-5.06 to 5.22	1.9	5.1
Peak Adduction	0.65	0.17 to 0.86	-3.3	-10.8 to 4.5	-0.31	-2.05 to 1.44	4.0	-8.21 to 7.6	2.9	7.9
Peak Internal Rotation	0.79	0.51 to 0.91	-13.1	-22 to -5.8	-0.48	-2.19 to 1.23	4.0	-8.23 to 7.28	2.8	7.8
Ankle Joint Angle (°)										
Peak Dorsiflexion	0.77	0.46 to 0.9	87.9	83.8 to 91.4	1.14	0.29 to 2	2.0	-2.72 to 5.01	1.4	3.9
Peak Abduction	0.85	0.64 to 0.93	20.4	3.6 to 33.3	-0.15	-2.53 to 2.23	5.5	-10.94 to 10.63	3.9	10.8
Peak External Rotation	0.82	0.59 to 0.93	24.7	16.8 to 30.7	-0.97	-2.33 to 0.4	3.2	-7.16 to 5.23	2.2	6.2
Peak Plantar Flexion	0.95	0.88 to 0.98	59.4	50.1 to 68.3	0.46	-0.48 to 1.4	2.2	-3.8 to 4.71	1.5	4.3
Peak Adduction	0.86	0.67 to 0.94	3.8	-9.5 to 18.4	-0.56	-2.39 to 1.27	4.2	-8.85 to 7.73	3.0	8.3
Peak Internal Rotation	0.86	0.66 to 0.94	14.9	5.8 to 21.2	-1.43	-2.72 to -0.14	3.0	-7.28 to 4.43	2.1	5.9

ICC, intraclass correlation coefficient; 95% CI, 95% confidence interval for the ICC; Mean, mean of measurements at time 1 and time 2; Mean (Min to Max), minimum and maximum mean value; D, mean of the differences between measurements at time 1 and 2 and the 95% CI for D; the standard deviation of the differences (SD_{diff}); 95% LOA, Bland and Altman 95% limits of agreement; SEM, standard error of measurement; MDC, minimal detectable change.

In order to better visualize the waveform agreement of joint angles, the mean and standard deviation of both assessments were plotted together (Figure 4). The largest curve agreement is in sagittal plane, except for thoracic segment, where higher variability can be seen in both assessment times. There is no tendency for major differences in overlapping and variability in key events of the gait cycle.

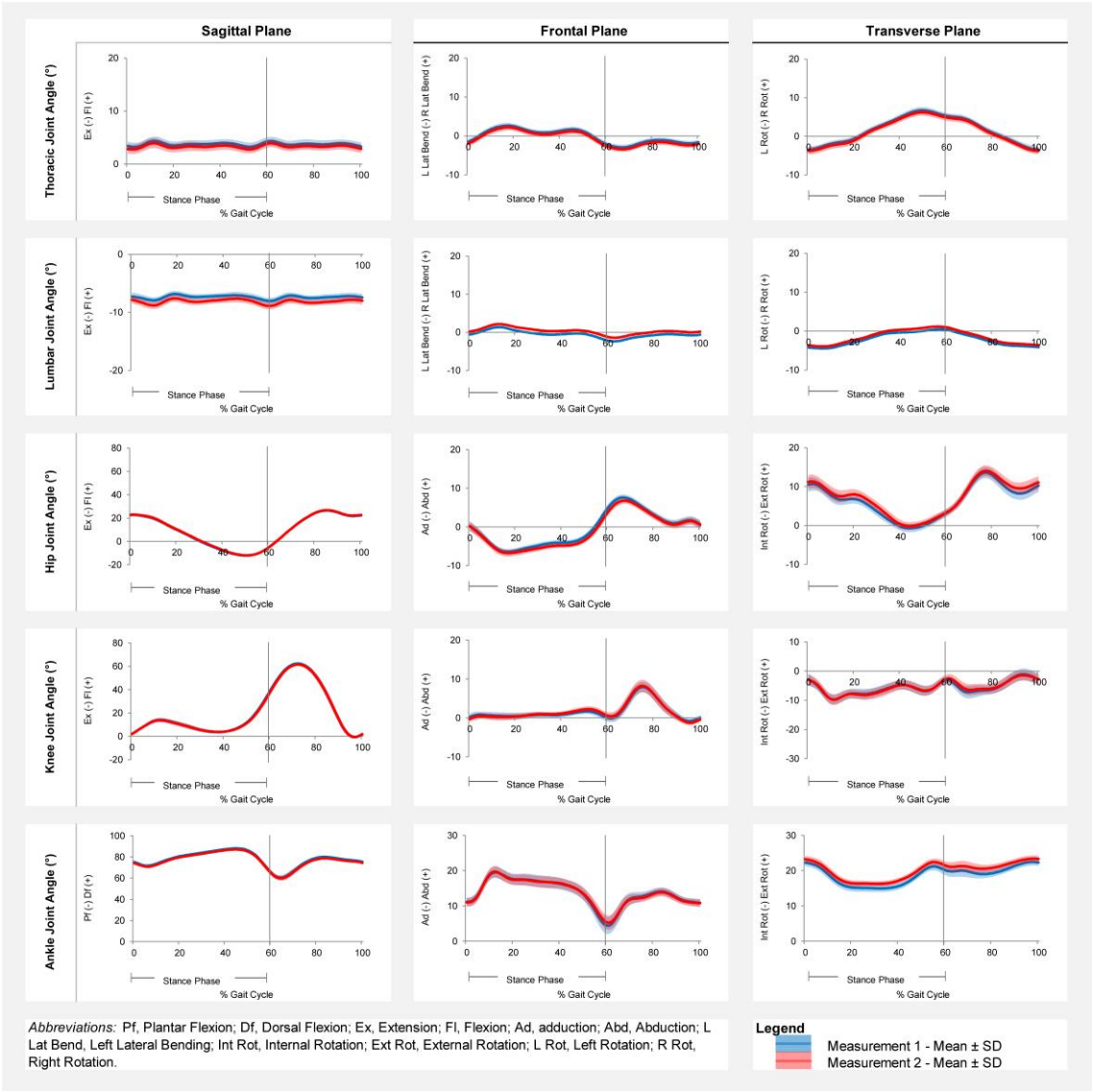

Figure 4 - Plots of joint angles waveforms during the gait cycle in the control group.

Reliability of kinetics was similar to kinematics, with almost half of the parameters showing ICCs ≥0.80 (Table 7). The remaining parameters showed ICCs between 0.53 and 0.80, with the exception of lumbar right lateral bending (0.19, 95% CI -1.0 to 0.67) and lumbar/thoracic right rotation (0.28, 95% CI -0.77 to 0.71 and 0.03, 95% CI -1.28 to 0.59, respectively). The SEM was ≤0.2 Nm/kg (varying from 0.02 to 0.2) and the MDC was ≤0.6 Nm/kg (varying from 0.1 to 0.5). Regarding waveform agreement, joint moments also showed the greatest differences and variability in frontal and transverse planes (Figure 5). There was a tendency for higher variability at gait cycle peaks (e.g. knee extension and hip flexion moments).

Table 7 - Reliability values for kinetic parameters in the control group.

Kinetic Parameter	ICC	(95% CI)	Mean	Mean (Min to Max)	D	(95% CI)	SD_{diff}	95% LOA	SEM	MDC
Thoracic Joint Moment (Nm/kg)										
Peak Flexion	0.84	0.63 - 0.93	0.36	0.1 to 0.7	0.04	-0.02 to 0.1	0.13	-0.22 to 0.3	0.09	0.26
Peak Right Lateral Bending	0.72	0.35 - 0.88	0.23	0 to 0.5	-0.02	-0.07 to 0.03	0.12	-0.25 to 0.21	0.08	0.23
Peak Left Rotation	0.85	0.63 - 0.93	0.09	0 to 0.2	0.01	-0.01 to 0.03	0.04	-0.07 to 0.09	0.03	0.08
Peak Extension	0.80	0.54 - 0.92	-0.24	-0.6 to 0.1	0.03	-0.04 to 0.09	0.14	-0.26 to 0.31	0.10	0.28
Peak Left Lateral Bending	0.64	0.16 - 0.85	-0.19	-0.7 to 0.1	-0.03	-0.1 to 0.04	0.16	-0.34 to 0.28	0.11	0.31
Peak Right Rotation	0.03	-1.28 - 0.59	-0.11	-0.2 to -0.1	-0.01	-0.03 to 0.02	0.06	-0.12 to 0.11	0.03	0.08
Lumbar Joint Moment (Nm/kg)										
Peak Flexion	0.87	0.67 - 0.95	0.42	0.2 to 0.7	0.03	-0.02 to 0.08	0.11	-0.19 to 0.25	0.15	0.41
Peak Right Lateral Bending	0.19	-1 - 0.67	0.19	0 to 0.3	0.05	-0.01 to 0.11	0.13	-0.21 to 0.31	0.10	0.26
Peak Left Rotation	0.75	0.38 - 0.9	0.07	0 to 0.2	-0.01	-0.02 to 0.01	0.03	-0.06 to 0.05	0.03	0.09
Peak Extension	0.87	0.68 - 0.95	-0.23	-0.4 to 0.1	-0.03	-0.08 to 0.03	0.12	-0.25 to 0.2	0.08	0.23
Peak Left Lateral Bending	0.66	0.16 - 0.86	-0.33	-0.7 to -0.1	0.01	-0.05 to 0.07	0.13	-0.25 to 0.27	0.16	0.45
Peak Right Rotation	0.28	-0.77 - 0.71	-0.13	-0.2 to -0.1	-0.02	-0.04 to 0	0.04	-0.1 to 0.07	0.04	0.11
Hip Joint Moment (Nm/kg)										
Peak Flexion	0.71	0.32 - 0.88	0.63	0.3 to 0.9	0.00	-0.07 to 0.07	0.16	-0.31 to 0.32	0.11	0.31
Peak Abduction	0.67	0.22 - 0.86	0.86	0.7 to 1.1	-0.03	-0.08 to 0.03	0.12	-0.26 to 0.21	0.08	0.24
Peak External Rotation	0.84	0.62 - 0.93	0.10	0 to 0.3	-0.01	-0.04 to 0.01	0.06	-0.12 to 0.1	0.04	0.11
Peak Extension	0.86	0.67 - 0.94	-0.39	-0.6 to -0.2	-0.02	-0.06 to 0.02	0.08	-0.18 to 0.15	0.06	0.17
Peak Adduction	0.53	-1.35 - 0.91	-0.09	-0.2 to 0	-0.01	-0.02 to 0.01	0.03	-0.06 to 0.04	0.02	0.05
Peak Internal Rotation	0.77	0.47 - 0.9	-0.17	-0.5 to -0.1	-0.03	-0.07 to 0.01	0.10	-0.22 to 0.16	0.07	0.19
Knee Joint Moment (Nm/kg)										
Peak Flexion	0.72	0.35 - 0.88	0.30	0.2 to 0.5	0.02	-0.02 to 0.06	0.09	-0.15 to 0.19	0.06	0.17
Peak Abduction	0.84	0.63 - 0.93	0.37	0.2 to 0.6	0.00	-0.03 to 0.04	0.09	-0.17 to 0.18	0.06	0.17
Peak External Rotation	0.60	0.07 - 0.83	0.09	0 to 0.1	0.00	-0.02 to 0.02	0.04	-0.08 to 0.08	0.03	0.08
Peak Extension	0.77	0.45 - 0.9	-0.54	-0.8 to -0.3	0.01	-0.06 to 0.07	0.14	-0.27 to 0.28	0.10	0.28
Peak Adduction	0.92	0.52 - 0.99	-0.04	-0.1 to 0	0.00	-0.01 to 0.01	0.02	-0.04 to 0.04	0.02	0.04
Peak Internal Rotation	0.52	-0.14 - 0.79	-0.13	-0.2 to -0.1	0.00	-0.02 to 0.02	0.05	-0.09 to 0.09	0.03	0.09
Ankle Joint Moment (Nm/kg)										
Peak Dorsiflexion	0.85	0.73 - 0.92	0.09	0 to 0.1	0.00	-0.02 to 0.01	0.03	-0.07 to 0.06	0.02	0.06
Peak Abduction	0.66	0.39 - 0.81	0.05	0 to 0.2	0.00	-0.03 to 0.02	0.06	-0.11 to 0.11	0.04	0.11
Peak External Rotation	0.86	0.76 - 0.92	0.11	0 to 0.2	0.00	-0.02 to 0.02	0.05	-0.09 to 0.09	0.03	0.09
Peak Plantar Flexion	0.98	0.96 - 0.99	-1.42	-1.7 to -1.3	-0.02	-0.07 to 0.03	0.12	-0.26 to 0.22	0.09	0.24
Peak Adduction	0.86	0.64 - 0.94	-0.31	-0.6 to -0.1	-0.01	-0.05 to 0.04	0.10	-0.21 to 0.2	0.07	0.20
Peak Internal Rotation	0.73	-0.07 - 0.93	-0.05	-0.1 to 0	0.00	-0.02 to 0.01	0.03	-0.07 to 0.06	0.02	0.07

ICC, intraclass correlation coefficient; 95% CI, 95% confidence interval for the ICC; Mean, mean of measurements at time 1 and time 2; Mean (Min to Max), minimum and maximum mean value; D, mean of the differences between measurements at time 1 and 2 and the 95% CI for D; the standard deviation of the differences (SD_{diff}); 95% LOA, Bland and Altman 95% limits of agreement; SEM, standard error of measurement; MDC, minimal detectable change.

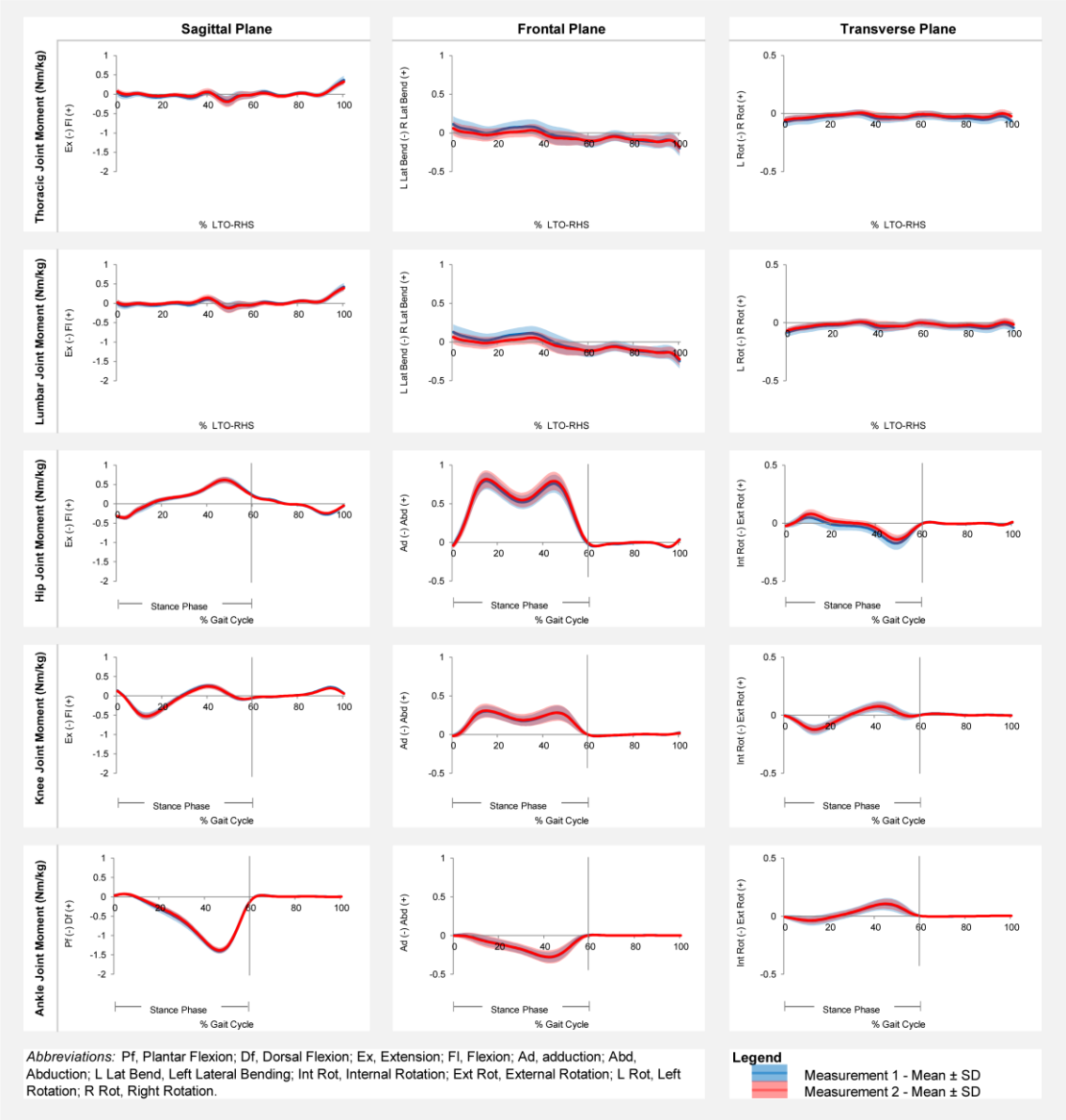

Figure 5 - Plots of joint moments waveforms during the gait cycle in control group.

4. Discussion

This is the first study that examined test-retest reliability and MDC of kinematic and kinetic 3DGA, considering thoracic and lumbar segments separately. The results suggest varied reliability indices for multi-segment trunk joint angles and joint moments, with acceptable reliability and level of error in the sagittal plane.

Our findings revealed that lumbar sagittal kinematics was reliable (ICCs of 0.91 and 0.90) and presented a measurement error around 3º. Despite the poorer reliability, SEM of frontal and transverse planes was low (1º to 3º) and the MDC ranged between 4º and 9º, indicating an absolute value for the amount of change that is sufficiently

greater than measurement error for lumbar segment in a healthy group (de Vet et al., 2006a). Thoracic segment revealed ICCs between 0.66 and 0.90 in sagittal/frontal planes and good agreement, with SEM values varying between 1º and 2º. Accordingly to McGinley et al. (2009), in most common clinical situations errors between 2º and 5º are likely to be regarded as reasonable but may require consideration in data interpretation. Nevertheless, since the variability of each parameter relative to the mean influences the interpretation of the measurement error, care should be taken when interpreting joint angles mainly based on absolute values of error. Accordingly, we looked into SEM percent change (SEM%) (Nair, Hornby, & Behrman, 2012) and we verified that despite the higher ICCs in lumbar peak of flexion/extension, moderate changes are needed to indicate a real change in both parameters (43% and 33%). Interestingly, thoracic right and left lateral bending showed a similar SEM% (27% and 28%), but significantly differed in ICC (0.90 and 0.77 for thoracic right and left lateral bending, respectively). These results support the importance of consider different issues when interpreting reliability and give an indication of the precision that can be expected when measuring kinematics of lumbar or thoracic segments.

The absence of data on test-retest reliability and measurement error of multi-segment trunk kinematics during gait in healthy individuals limits comparisons with standard data regarding this population. Nonetheless, these results are consistent with those of Wilken et al. (2012) who computed the absolute angle of the trunk in a sample of healthy individuals and reported SEM and MDC values for frontal and sagittal planes below 2º. These authors did not provide the mean value of each parameter, which prevents any interpretation regarding variability and mean.

Almost half of the lower limb joint angles (obtained from averages of 10 gait cycles) showed ICCs greater than 0.80 (0.81 to 0.95), with the remaining showing values between 0.60 and 0.80. These results are consistent with Meldrum et al. (2014) nevertheless they seem to be more repeatable for hip, knee and ankle frontal/transverse planes peak parameters. In line with McGinley et al. (2009), reliability of transverse and frontal planes was generally lower (median of 0.70 and 0.76) than sagittal plane (median of 0.81), nonetheless the diversity of study types, participants and methodologies limits between studies comparisons. The peaks of knee extension/adduction and hip abduction/external rotation angles were the lower limb parameters with poorer reliability (ICCs from 0.64 to 0.69), however when interpreted alongside their SEMs, these parameters showed a measurement error between 2º and 5º, which agrees with the commonly reported in gait studies (McGinley et al., 2009). Considering the SEM%, our results suggest that significant changes are needed to

indicate a real change in knee extension (122%) and adduction (86%), while smaller changes are needed to indicate a real change in hip abduction (22%) and external rotation (33%). This may also suggest that these hip parameters may have an acceptable level of error and their use can be considered, contrary to knee joint.

No systematic bias was detected for trunk or lower limb kinematics and there were no indications of heteroscedasticity, which reinforces the agreement of the measurements. Nevertheless, the 95% LOA intervals of some parameters (particularly those of transverse plane) were relatively wide, indicating that a substantial difference in an individual joint angle would be required to allow us to confidently state that a real individual change had taken place (Mieritz, Bronfort, Jakobsen, Aagaard, & Hartvigsen, 2014).

The reliability of joint moment peaks was similar to joint angles, with almost half of the parameters showing ICCs ≥0.80 (0.80 to 0.98). The remaining ICCs fell between 0.53 and 0.80, excepting lumbar right lateral bending (0.19, 95% CI -1.0 to 0.67) and lumbar/thoracic right rotation (0.28, 95% CI -0.77 to 0.71 and 0.03, 95% CI -1.28 to 0.59, respectively), which revealed poorer reliability. These results agree with Wilken et al. (2012) and are slightly better than Meldrum et al. (2014), however comparisons with those studies are limited since the authors did not included information regarding trunk kinetics. Joint moments indicated lower SEM% comparatively to joint angles, suggesting that a lower change is needed to indicate a real change. Specifically, only ankle abduction/internal rotation, lumbar right lateral bending/left rotation and thoracic extension/left lateral bending revealed a SEM% superior to 40%. The 95% LOA intervals for joint moments in sagittal plane were narrow and the curve overlap between assessments was higher, supporting their agreement and use in clinical and research.

Strengths and limitations:

One of the strengths of this study is that the 2 measurements were separated by a mean of 7 days, with all evaluations occurring at the same time of the day to diminish interference of recall bias. Since the assessor was blinded to the results of the first assessment, his/her interference was limited. Testing conditions also contributed to high reliability of our measurements, with participants allowed to adopt their normal gait (Monaghan et al., 2007) and a very stable pace. There are also a few limitations in this study. Firstly, lumbar and thoracic segments were assumed to be single rigid segments. This is a highly simplified model of spinal motion, however it might be appropriate as a first step for developing a better understanding of reliability of spinal

motion during gait. Some parameters showed wide 95% CIs for ICCs, which may contribute to some uncertainty in the findings. The influence of gender differences has been tested. However, relevant reliability was not supported by groups' size and the large 95% CIs around reliability indices limits any definitive conclusion about gender dependence. Therefore, we recommend the investigation of separate groups in future reliability studies.

3.5 Conclusions

The results of this study suggest varied reliability indices for multi-segment trunk joint angles and joint moments during gait and an acceptable level of error, particularly for sagittal plane parameters. Some parameters showed wide 95% CIs for ICCs and higher SEM%, which may contribute to some uncertainty in the findings. However, we believe this study provide preliminary data regarding reliability indices for multi-segment trunk during gait, which may be valuable for clinical reasoning and decision making when dealing with musculoskeletal disorders.

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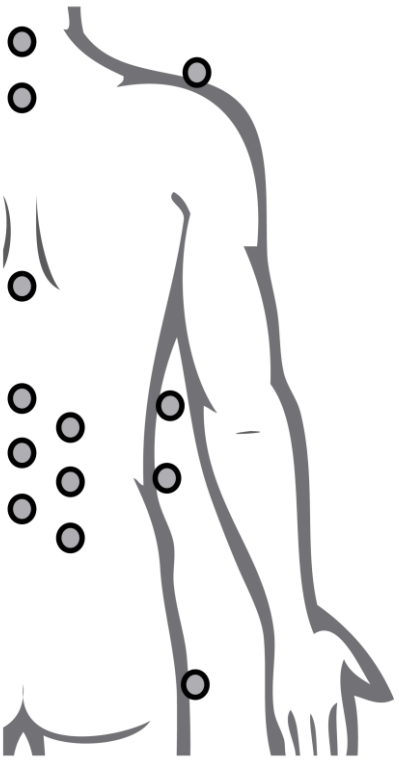
4

Loss of variability and altered three-dimensional trunk and hip kinetics during gait in chronic low back pain individuals



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Abstract

Background and Aim: Combining information on kinetics and kinematics of the trunk during gait is important, since it can help in better understanding the mechanisms behind changes in movement patterns in CLBP. This study aims at determining the differences between CLBP and healthy individuals in kinematics and kinetics of thoracic, lumbar and hips during gait, taking into account the error values; and to gain insight into the variability of movement between the mentioned segments in association with joint moments.

Methods: Nineteen CLBP patients and twenty controls walked on barefoot at their preferred speed. Time-distance parameters and joint angles/moments peaks were computed. Step-to-step variability of thoracic, lumbar and hip segments was calculated and correlated to each other.

Results: In CLBP individuals, the sagittal and transverse planes residual rotations of lumbar were positively correlated in magnitude and negatively correlated in sign with sagittal and frontal planes residual rotations of thoracic segment. A decrease in lumbar/thoracic flexor joint moments and an increase in thoracic axial joint moment were verified in patients.

Conclusions: Stride-to-stride variability of lumbar and thoracic segments is significantly and inversely related in CLBP individuals with respect to controls, supporting the argument that patients adopt a protective movement strategy.

Keywords

Trunk-pelvis coordination; residual rotation; joint moments; joint angles.

4.1 Introduction

Motor control is a key component of efficient movement patterns and in daily activities where precise motor control is required (i.e., standing, balancing, or gait) a more flexible spine may be beneficial (Reeves, Narendra, & Cholewicki, 2007). Healthy individuals compensate for internal and external perturbations that potentially disrupt gait, showing a flexible reorganization that preserves stable gait patterns (Lamoth, Daffertshofer, Meijer, & Beek, 2006). This adaptive response is possible due to the high degree of coordination between cyclically moving body segments (e.g. limbs, pelvis, trunk, and head) that characterises unimpaired gait (Lamoth et al., 2006). These interactions or couplings are relatively stable, yet able to adapt to changes, being the variation in walking speed a relevant one (Lamoth et al., 2006). In the last years, research in CLBP individuals has often focused on the variability and coordination of trunk and pelvis kinematics during gait. Although conflicting, findings suggest that CLBP individuals exhibit a reduced ability to adapt trunk–pelvis coordination in response to changes in gait velocity (Lamoth et al., 2006), display a more rigid pelvis-thorax coordination (Lamoth et al., 2002) and have lower variability of trunk rotations, as a result of the coupling of deviations of residual rotations (in shape and amplitude) between pelvis and trunk (van den Hoorn, Bruijn, Meijer, Hodges, & van Dieën, 2012). Particular attention has been given to the transverse plane rotations during gait, with CLBP individuals showing difficulty in moving from pelvis-trunk in-phase (synchronous rotations in the same direction) to anti-phase rotations (synchronous rotations in opposite direction) as gait velocity increases, which is the observed response/pattern in healthy individuals (Lamoth et al., 2006). This phase difference between two oscillating segments is closely related to the clinical evaluation of movement patterns, which is an important part of the evaluation process and treatment selection when dealing with CLBP individuals (Lamoth et al., 2002).

The kinematic analysis of functional activities is highly valuable, however, it remains descriptive and cannot fully explore the biomechanical mechanisms underlying changes in movement strategies and the nature of the loading patterns in the lumbar spine (Shum, Crosbie, & Lee, 2007). Previous studies that attempted to estimate kinetic variables in LBP individuals have mainly focused on functional activities that included flexion and extension of the trunk, namely lifting tasks (Kingma et al., 2001; Marras, Davis, Ferguson, Lucas, & Gupta, 2001), sit-to-stand and reverse (Shum, Crosbie, & Lee, 2009; Shum et al., 2007), as well as backward/forward bending (Shum, Crosbie, & Lee, 2010). Those studies found that, compared to healthy subjects, LBP

individuals had decreased sagittal joint moments acting on the lumbar spine at the end of the available range during forward/backward bending and sit-to-stand, but had increased axial joint moments during sit-to-stand. Previous findings also showed that LBP individuals recruit their muscles differently and have altered flexion-relaxation responses (Alschuler, Neblett, Wiggert, Haig, & Geisser, 2009), suggesting distinctive muscle activation patterns that may impose an altered load on the lumbar spine (Shum et al., 2007).

According to our best knowledge, studies with CLBP individuals that focused on complex activities, as gait, have limited their analysis to kinematic and electromyographic variables (Gombatto et al., 2015; Lamoth et al., 2006; van den Hoorn et al., 2012; Vogt, Pfeifer, Portscher And, & Banzer, 2001). Additionally, kinematic data have been interpreted and discussed without taking into account their error magnitude, which can maximize the risk of over-interpreting small differences between groups as meaningful (Fernandes, Armada-da-Silva, Pool-Goudzwaard, Moniz-Pereira, & Veloso, 2015; McGinley, Baker, Wolfe, & Morris, 2009). This is even more important in gait analysis, since we already know that error values associated with this procedure are generally not small enough to be ignored during the interpretation of clinical data (McGinley et al., 2009). Further, combining the information on kinematics with multi-trunk kinetics during gait is of importance, since it can offer a deeper understanding about the causes of movement pattern changes in CLBP individuals. Thus the aims of this study are: 1) to determine differences in thoracic, lumbar and hip kinematics and kinetics between CLBP patients and healthy individuals during gait, taking into account the error values; 2) and to gain insight into the variability of movement between thoracic, lumbar and hip segments in association with joint moments, in CLBP patients versus healthy individuals.

4.2 Materials and Methods

4.2.1 Study Design

A cross-sectional study was conducted.

4.2.2 Participants

In the absence of a clear primary outcome regarding biomechanical parameters and based on the most usual sample sizes, a convenience sample of 23 CLBP individuals and 26 healthy volunteers was respectively recruited from community/outpatient clinics and university staff/associates, according to a standardized recruitment protocol.

Physiotherapists from the research team and outpatient clinics carried out patient recruitment based on predefined inclusion/exclusion criteria. Patients were considered eligible if they were aged between 18 and 65 years, and had LBP, with or without referred leg pain, for at least 12 weeks (Airaksinen et al., 2006) or recurrent LBP (Von Korff, 1994). Eligible patients were screened for evidence of serious low back pain pathology and were excluded if they had clinical signs of infection, tumour, osteoporosis, fracture, structural deformity, inflammatory disorder (e.g. ankylosing spondylitis), radicular syndrome, cauda equine syndrome, or if they had undergone back or lower limb surgery or a conservative treatment in the prior 12 and 6 months, respectively. Healthy individuals were considered eligible if they were aged between 18 and 65 years old and were excluded if they had any clinical condition (musculoskeletal, neurological, cardiac or pulmonary) or symptom that could affect gait. Pregnant women were excluded from both groups. After this screening, 19 of the 23 CLBP patients and 20 of the 26 healthy individuals were included in the study.

The local Ethics Committee approved the study. All the participants were informed of the procedures and risks of the study and signed an informed consent.

4.2.3 Procedures

Before testing, participants' clinical history was reviewed and a standard physical examination focussed on lumbar spine and lower limbs was performed. This was complemented with the measurement of body mass and height. Segments' length was obtained using the respective proximal and distal anatomical landmarks collected during the static trial described below. For pelvis, ASIS and PSIS markers were used. In order to complete CLBP individuals' clinical characteristics, pain intensity, disability and kinesiophobia were assessed using the NRS, the QBPDS-PT (Cruz et al., 2013) and Tampa Scale of Kinesiophobia (TSK-PT) (Cordeiro, Pezarat-Correia, Gil, & Cabri, 2013), respectively. The Baecke Physical Activity Questionnaire (BPAQ) assessed the physical activity level of participants in both groups.

Finally, gait data was collected using a 13-camera opto-electronic system (Oqus 300, Qualisys AB, Gothenburg, Sweden) synchronized in time and space with two Kistler (9281B and 9283U014, Kistler Group, Winterthur, Switzerland) and one AMTI (BP6001200, Advanced Mechanical Technology, Inc Watertown, USA) force platforms at 200Hz. The marker set used was based on previous reports (Leardini, Biagi, Merlo, Belvedere, & Benedetti, 2011; Seay, Selbie, & Hamill, 2008) (Figure 1). After a static trial, participants were instructed to walk barefoot at their preferred walking speed, continuously in a 14-meter walkway, and during short periods of time (1-2 minutes) to

avoid fatigue. A familiarization trial was performed before data collection. The same assessor carried out all the procedures.

4.2.4 Data Processing

Considering the natural variability in kinematic and kinetic gait parameters, 10 cycles were selected (Monaghan, Delahunt, & Caulfield, 2007). Cycles were extracted using Qualysis Track Manager (v2.8 build 1554, Qualisys AB, Gothenburg, Sweden) and exported to be processed under Visual 3D software (v5.01.10, C-Motion, Inc, Rockville, USA).

A 9-segment model (feet, shanks, thighs, pelvis, lumbar and thoracic spine) was built for each participant (Leardini et al., 2011; Seay et al., 2008). All the LCSs were defined in accordance with Robertson, Caldwell, Hamill, Kamen, & Whittlesey (2014) and their origin was the joint centre. The ankle and knee joint centres were defined as the midpoint of the tibia malleoli and as the midpoint of the femur epicondyles, respectively (Robertson et al., 2014). The hip joint centres were computed using the pelvis markers, according to published regression equations (Bell, Pedersen, & Brand, 1990). The lumbar joint centre was defined through a virtual marker created along the distance connecting the L5-S1 marker and the midpoint between the two ASIS markers (Seay et al., 2008), projected from the thoracic joint centre. The thoracic joint centre was defined using a virtual marker projected from the midpoint of the markers placed bilaterally on the ribcage at the T12-L1 joint space level onto the thorax longitudinal axis. The proximal end of this axis was defined as the midpoint between the suprasternal notch and the second thoracic vertebra, while the distal end was defined as the midpoint between the xiphoid process and the inferior angles of most caudal points of the two scapulae. At the pelvis, two markers were placed on each anterior and posterior superior iliac spine, as well as on top of each iliac crest. The proximal end was defined using the virtual marker created as the distal end of the lumbar segment and the distal end was defined as the midpoint between the hips. A second LCS (for kinematic computations only) was created based on the CODA pelvis model (Robertson et al., 2014) in order to achieve a more clinically recognisable pelvic tilt (sagittal plane). Each segment was considered to be independent and to have 6 degrees of freedom (SO method) (Cappello, La Palombara, & Leardini, 1996). Lower limb segment masses were determined according to Dempster (1955), while the remaining inertial parameters were computed based on Hanavan (1964). Lumbar and thoracic inertial parameters were computed according to Pearsall, Reid and Livingston (1996).

A Woltring generalized cross-validatory cubic spline smoothing routine (Woltring, 1986) with an error variance of 0.0001 (mean standard error of 1 mm) was used to filter kinematic and kinetic data. Lower limb and trunk joint angles (using a XYZ Cardan sequence) and joint moments (determined through inverse dynamics and normalized to subjects' body mass) were computed consistent with Robertson et al. (2014; pp. 50-54 and 152-164) and expressed relatively to the proximal segment. Thus, flexion/extension rotations occurred around the medio-lateral axis of the proximal segment, abduction/adduction/lateral bending rotations around a floating axis and external/internal/right/left rotations around the distal segment longitudinal axis. Data was normalized to 100% stride cycle, time-distance parameters were normalized to subjects' height and joint moments to subjects' mass. Additionally peak values for hip and trunk joint angles and joint moments, as well as time-distance parameters, were computed for each cycle and averaged for each subject.

4.2.5 Data Analysis

After testing for variables normal distribution using the Kolmogorov-Smirnov test, independent samples t-tests and chi-square tests were used to test for differences in participants' characteristics and gait variables (time-distance parameters and peak values for trunk and hips joint angles and moments).

Stride-to-stride variability was calculated based on van den Hoorn et al. (2012). The average stride cycle (1x101) was subtracted from each stride cycle, which resulted in a matrix of strides x 101 residual rotations for each time series. These amplitudes (in degrees) represented the difference from the average rotational component, and could be positive and negative throughout the stride cycle. The mean absolute residual rotations of thoracic, lumbar, and hips were calculated over all strides resulting in three 1x101 vectors. The maximum, minimum and median values of each vector were calculated to represent variability (in degrees) of the mentioned segments. After testing for residuals rotations normal distribution, spearman correlations were calculated to assess the relationship between the residual rotations (signed both negative and positive) of thoracic, lumbar and hips. When significant, a minimum of 0.50 was considered to be an acceptable correlation coefficient, as it represents a moderate correlation between two different variables.

The statistical analysis was conducted using SPSS (version 20.0; IBM, Chicago, IL) and a critical level of $p<0.05$ was considered significant. Joint angles/moments waveforms were plotted using Microsoft Excel 2007 (Microsoft Corp., Redmond, WA).

4.3 Results

There were no significant differences in subjects' characteristics (Table 8) or gait parameters (Table 9) between groups. Left hip was chosen to present the results, as no statistical differences were observed between the right and left hips in groups' comparison.

Table 8 - Subjects characteristics' and self-report measures in the pain and control group.

	Control Group (n=20)	**Pain Group** (n=19)	***p* value**
Characteristics			
Age (yr)	42.2 (7.95)	47.05 (7.9)	0.06
Gender (n)	F=12; M=8	F=14; M=5	0.30
Height (cm)	1.68 (0.07)	1.65 (0.08)	0.17
Mass (kg)	66.4 (11.11)	65.92 (11.84)	0.89
BMI (Kg/m^2)	23.5 (2.59)	24.25 (3.34)	0.43
Self-Report Measures			
BAECKE - Work Index	2.38 (1.75 - 3.25)	2.38 (1.88 – 3.00)	0.63
BAECKE - Sport Index	2.2 (0.75 - 5.53)	3.22 (1.00 - 4.55)	0.12
BAECKE - Leisure Index	2.75 (2.00 - 3.75)	3.00 (2.00 - 4.50)	0.29
QBPDS-PT (0-100)	-	20.00 (3.00 – 38.00)	-
NRS – 24 hours (0-10)	-	2.00 (0 – 7.00)	-
NRS - 1 week (0-10)	-	3.00 (1.00 – 9.00)	-
TSK-PT (0-52)	-	28.00 (20.00 – 40.00)	-

Data is presented as mean (standard deviation) for subjects characteristics' and median (max-min) for self-report measures.

Table 9 - Gait parameters in the pain and control group.

Temporal Distance Parameter	**Control Group** (n=20)	**Pain Group** (n=19)	***p* value**
Speed (m/s)	1.19 (0.12)	1.16 (0.20)	0.58
Speed (m/s/height)	0.72 (0.10)	0.71 (0.14)	0.90
Cycle Time (s)	1.08 (0.09)	1.07 (0.10)	0.88
Double Limb Support Time (s)	0.11 (0.02)	0.13 (0.04)	0.06
Stride Length (m)	1.28 (0.09)	1.23 (0.12)	0.13
Stride Length (m/height)	0.76 (0.06)	0.75 (0.08)	0.51
Stride Width (m)	0.09 (0.02)	0.09 (0.02)	0.81
Stride Width (m/height)	0.08 (0.02)	0.08 (0.03)	0.76

Data is presented as mean (standard deviation).

Regarding trunk kinematic parameters, there were no significant differences between CLBP individuals and controls (Figure 6), however, both groups significantly differed in the peak of hip abduction/adduction ($p<0.02$) and internal/external rotation ($p<0.01$), but not in the range of motion (Figure 7).

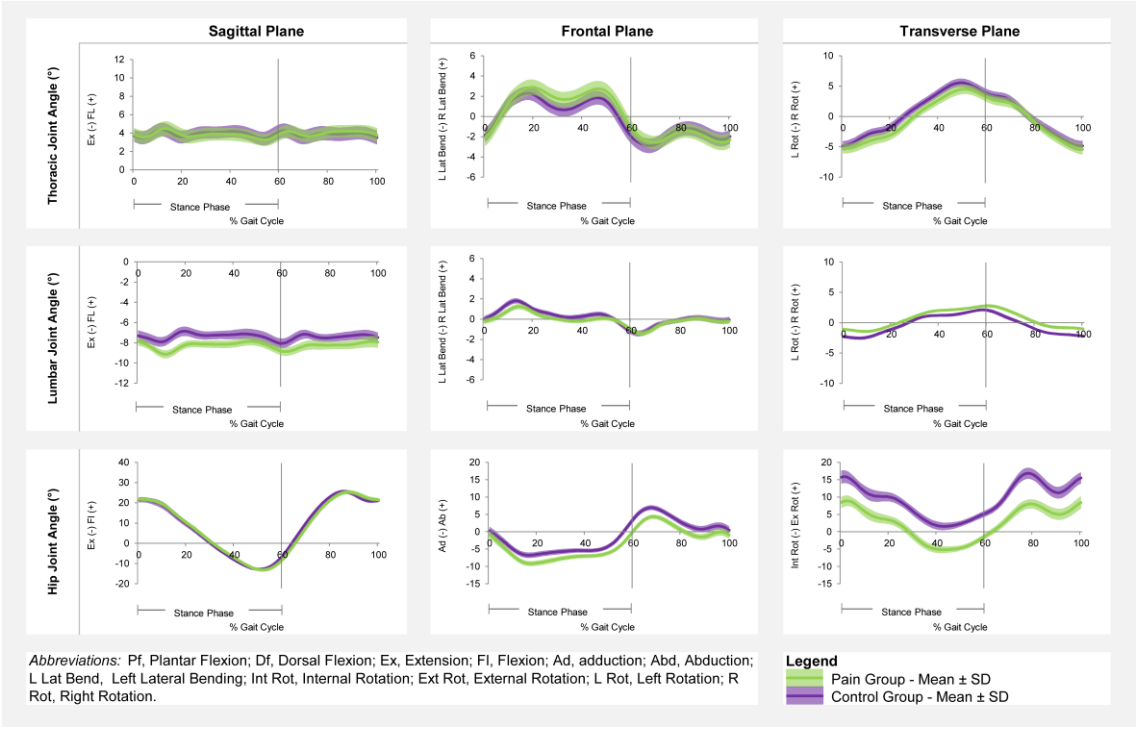

Figure 6 - Thoracic, lumbar and hip joint angles gait cycle waveforms (sagittal, frontal and transverse planes) in the pain and control group.

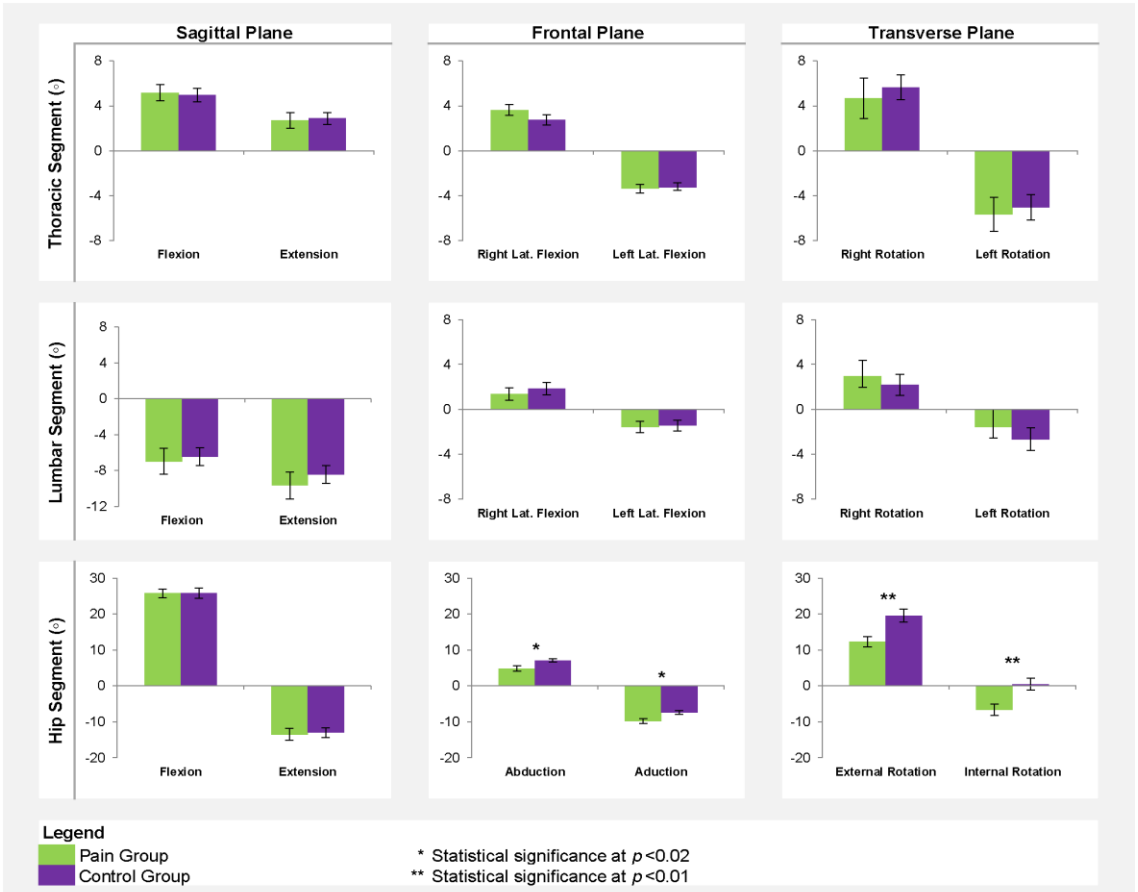

Figure 7 - Mean (SE) differences in the peaks of joint angles of the thoracic, lumbar and hip segments during gait in the pain and control group.

Contrary to kinematics, trunk sagittal plane kinetics significantly differed between groups (Figure 8). Specifically, CLBP individuals showed lower lumbar flexor joint moment (0.20 and 0.41 Nm/kg respectively, $p<0.01$) and thoracic flexor joint moment peaks (0.27 and 0.39 Nm/kg respectively, $p<0.04$), comparatively to control individuals (Figure 9). Thoracic rotation joint moment peak was also significantly different between groups, however, contrary to sagittal plane, it was higher in CLBP (0.09 Nm/kg) compared to control group (0.07 Nm/kg), for $p<0.04$.

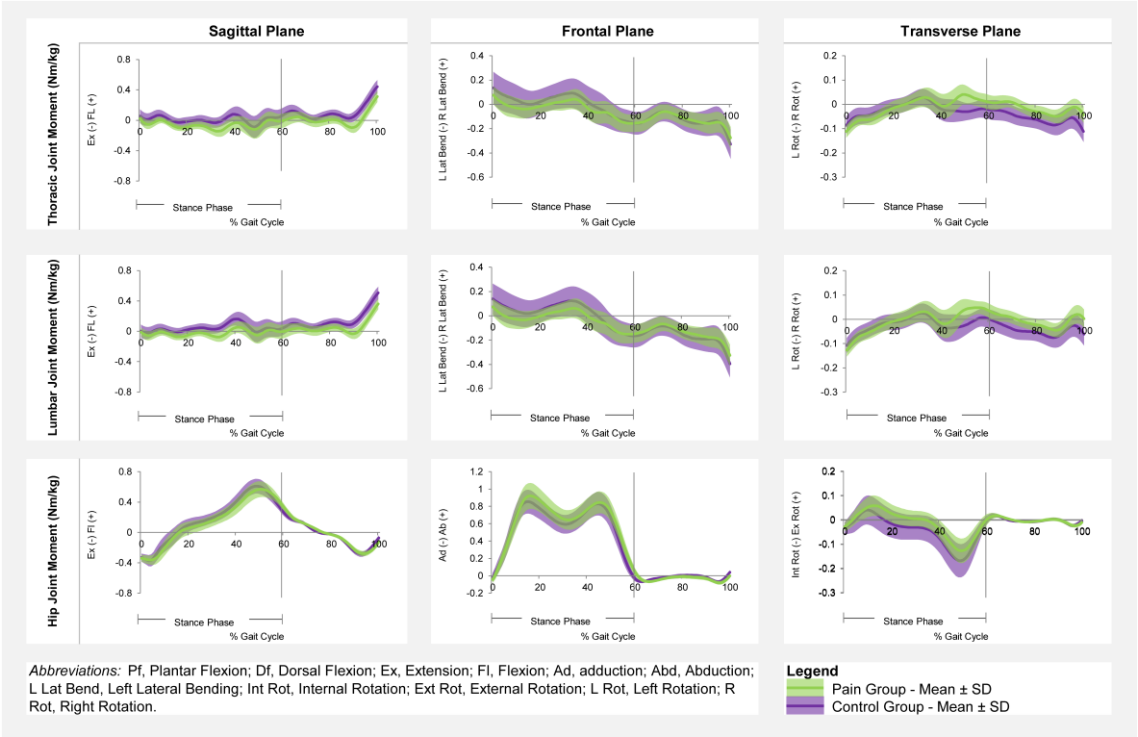

Figure 8 - Thoracic, lumbar and hip joint moments gait cycle waveforms (sagittal, frontal and transverse planes) in pain and control group.

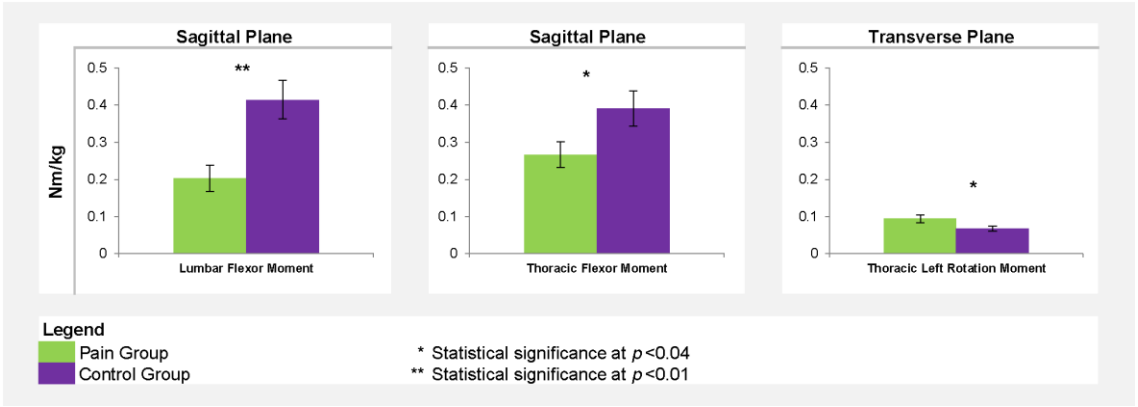

Figure 9 - Mean (SE) differences in the peaks of joint moments of the thoracic, lumbar and hip segments during gait in the pain and control group.

Concerning variability, the mean absolute residual rotations of thoracic, lumbar, and hip segments did not differ between groups ($p<0.05$). This result is consistent with the absence of significant differences between groups with respect to the maximum, minimum and median values of residual rotations vectors ($p>0.05$). Even so, it is possible to identify a tendency for higher dispersion in residual rotations correlations within control group (Figure 10).

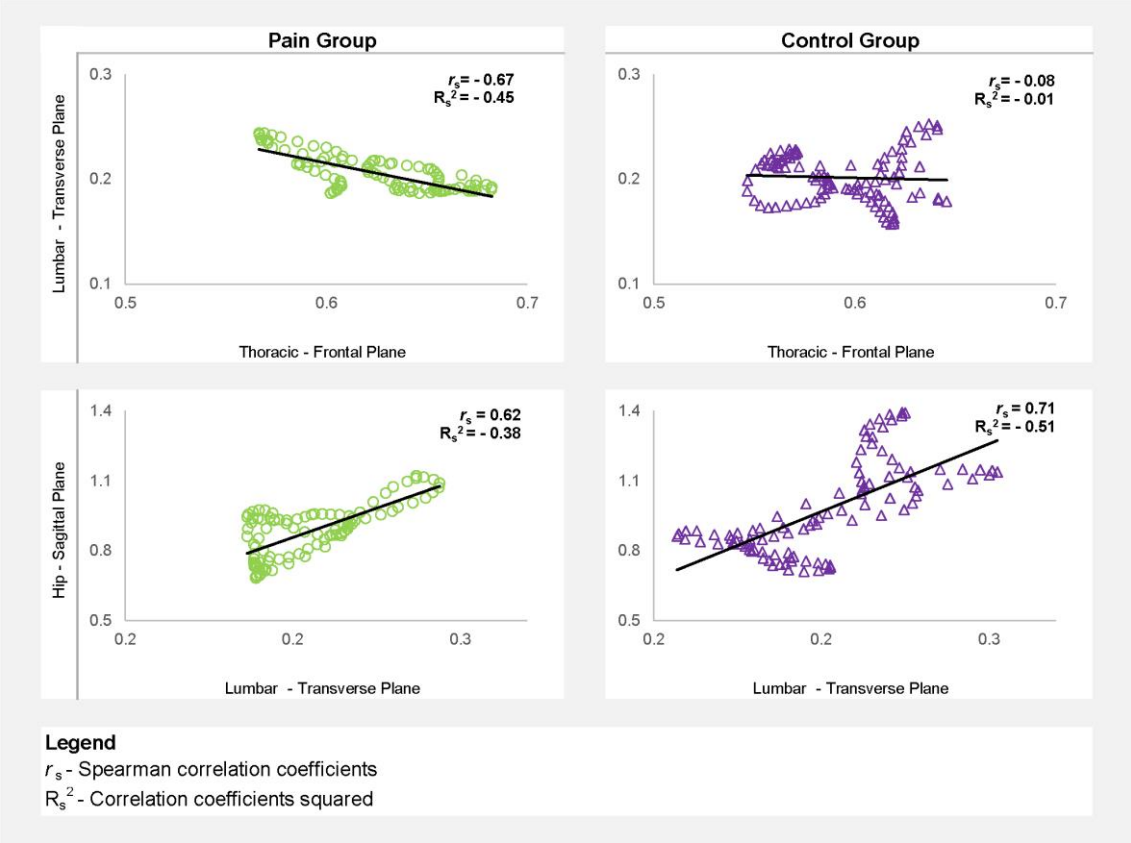

Figure 10 - Residual rotations (º) of the thoracic (frontal) in x-axis and lumbar (transverse) in y-axis and residual rotations of the lumbar (transverse) in x-axis and hip (sagittal) in y-axis.

Spearman correlations between residual rotations of thoracic and lumbar significantly differed between CLBP and healthy individuals (Table 10). After selecting the significant correlation coefficients, we verified that the residual rotations of the lumbar and thoracic segments were differently correlated in amplitude and sign, in both groups. Specifically, in the CLBP group, the sagittal and transverse planes residual rotations of lumbar segment were correlated in magnitude and opposite sign with the sagittal and frontal planes residual rotations of thoracic segment (-0.66 and -0.53, p=0.000), which was not verified in the control group ($p>0.05$). This indicates that higher stride-to-stride variability in sagittal and transverse planes of lumbar segment was correlated with lower variability in sagittal and frontal planes of the thoracic segment in CLBP group (and vice versa); while in control group this relation was the

opposite, had a low magnitude or was not significant. The transverse plane residual rotations of the thoracic segment show the same pattern in both groups (p<0.02); however, the magnitude of the correlation with lumbar transverse plane residual rotations was clearly higher in the CLBP group (-0.59, *p*=0.000).

Table 10 - Spearman correlation coefficients (r_s), *p* values and correlation coefficients squared (R_s^2) from thoracic, lumbar and hip residual rotations of the pain and control group.

		Control Group			Pain Group		
		r_s	*p*	R_s^2	r_s	*p*	R_s^2
Lumbar Flexion/Extension	Lumbar Lat. Flexion	**.50****	.000	**0.25**	**.23***	.021	0.05
	Lumbar Axial Rotation	.16	.115	0.03	**.22***	.025	0.05
	Thoracic Flexion/Extension	.08	.449	0.01	**-.66****	.000	**0.44**
	Thoracic Lat. Flexion	.18	.069	0.03	**-.53****	.000	**0.28**
	Thoracic Axial Rotation	**.22***	.027	0.05	**.30****	.002	0.09
	Hip Flexion/Extension	-.07	.466	0.01	.18	.065	0.03
	Hip Abduction/Adduction	.08	.428	0.01	**.35****	.000	0.12
	Hip Internal/External Rotation	**-.26****	.010	0.07	**.20***	.043	0.04
Lumbar Lateral Flexion	Lumbar Axial Rotation	.01	.956	0.00	-.02	.867	0.00
	Thoracic Flexion/Extension	**.50****	.000	**0.25**	**.28****	.004	0.08
	Thoracic Lat. Flexion	**.64****	.000	**0.40**	-.10	.317	0.01
	Thoracic Axial Rotation	**.68****	.000	**0.46**	**.47****	.000	0.22
	Hip Flexion/Extension	.04	.697	0.00	**.47****	.000	0.22
	Hip Abduction/Adduction	**.62****	.000	**0.38**	**.46****	.000	0.21
	Hip Internal/External Rotation	**.27****	.007	0.07	**.27****	.007	0.07
Lumbar Axial Rotation	Thoracic Flexion/Extension	**.33****	.001	0.11	**-.36****	.000	0.13
	Thoracic Lat. Flexion	-.08	.434	0.01	**-.67****	.000	**0.45**
	Thoracic Axial Rotation	**-.34****	.000	0.12	**-.59****	.000	**0.35**
	Hip Flexion/Extension	**.71****	.000	**0.51**	**.62****	.000	**0.38**
	Hip Abduction/Adduction	-.15	.123	0.02	-.14	.175	0.02
	Hip Internal/External Rotation	-.18	.080	0.03	-.19	.065	0.03
Thoracic Flexion/ Extension	Thoracic Lat. Flexion	**.66****	.000	**0.44**	**.53****	.000	**0.28**
	Thoracic Axial Rotation	**.42****	.000	0.18	**.23***	.019	0.06
	Hip Flexion/Extension	**.27****	.007	0.07	.07	.517	0.00
	Hip Abduction/Adduction	.17	.096	0.03	.00	.987	0.00
	Hip Internal/External Rotation	-.02	.834	0.00	-.04	.731	0.00
Thoracic Lateral Flexion	Thoracic Axial Rotation	**.57****	.000	**0.33**	**.23***	.022	0.05
	Hip Flexion/Extension	-.17	.099	0.03	**-.61****	.000	**0.38**
	Hip Abduction/Adduction	.17	.100	0.03	-.09	.325	0.01
	Hip Internal/External Rotation	-.01	.974	0.00	-.18	.071	0.03
Thoracic Axial Rotation	Hip Flexion/Extension	**-.29****	.003	0.09	-.13	.181	0.02
	Hip Abduction/Adduction	**.71****	.000	**0.51**	.19	.061	0.04
	Hip Internal/External Rotation	**.51****	.000	**0.26**	.07	.493	0.01

The results also show that the proportion of variance shared by the ranks of the residual rotations of lumbar and thoracic transverse plane is 35% (Rs2 = 0.35). In both groups there was a significant positive correlation between the frontal planes of the lumbar and hip residual rotations (0.68 and 0.47, *p*=0.000) and between lumbar transverse plane and hip sagittal plane residual rotations (0.71 and 0.62, *p*=0.000).

4.4 Discussion

In this study we determined the differences between CLBP and healthy individuals in the kinematics and kinetics of thoracic, lumbar and hip during gait, taking into account the error values, and gained insight into the variability of movement between thoracic, lumbar and hip segments in association with joint moments, in both groups. As far as we know, this is the first study comparing multi-trunk kinematics and kinetics during gait in these groups. Our results suggest that stride-to-stride variability of lumbar and thoracic segments is significantly and inversely related in CLBP individuals, compared to controls. In CLBP group, greater variability in sagittal and transverse planes of lumbar segment is correlated with lower variability in sagittal and frontal planes of the thoracic segment (and vice versa), while in the control group, when significant, higher variability in one segment is mostly correlated with higher variability in the other. The results also showed that there is a decrease in the lumbar and thoracic flexor joint moments, but an increase in the thoracic axial joint moment during gait, in CLBP individuals. These differences are out of the established SEM values (Fernandes et al., 2015; Fernandes, Armada-da-Silva, Pool-Goudzwaard, Moniz-Pereira, & Veloso, 2016), indicating that trunk joint moments results are not masked by the measurement error.

Literature has consistently pointed out that stiffness of the trunk is increased in LBP patients during gait, which seems to be related with the reduced variability of trunk motion (Lamoth et al., 2006; van den Hoorn et al., 2012). These findings suggest a protective strategy of LBP individuals to avoid painful motion in order to prevent recurrence or pain provocation (van Dieën, Selen, & Cholewicki, 2003). In this study we found a different relationship between variability of the lumbar and thoracic segments, in CLBP and healthy individuals. Greater variability in lumbar sagittal and transverse planes was significantly correlated with lower variability in thoracic sagittal and frontal planes (and vice versa), suggesting increased stiffness in the thoracic region as a protective strategy in response to higher segmental variability in the lower trunk region exhibited by CLBP individuals. Our findings are in line with Lamoth et al. (2006, 2002), who verified that the significant difference between CLBP and healthy individuals was

in the intersegmental coordination between the rotations of the pelvis and thorax, particularly in response to changes in gait velocity, and not in the amplitudes of the individual rotations. These authors verified that CLBP individuals show a reduced ability in moving from pelvis-trunk in-phase (pelvis and trunk moving in the same directions) to anti-phase (pelvis and trunk moving in opposite directions) as walking speed increases (Lamoth et al., 2006). Similarly, van den Hoorn et al. (2012) found that individuals with CLBP had lower variability of trunk rotations, as a result of the deviations coupling of residuals rotations (in shape and amplitude) between pelvis and trunk. Our results appear indicate that adjacent trunk segments may show different stride-to-stride variability during gait in CLBP individuals, which may suggest that increasing stiffness in one part of a system does not always increase the overall system's stiffness (Reeves et al., 2007). Although literature has strongly pointed out that CLBP individuals' tend to move their lumbar and pelvic segments as a rigid unit and present a less flexible pelvis-thorax coordination (Lamoth et al., 2002), our findings regarding the significant differences in variability between two adjacent trunk segments has not been reported previously.

The results regarding kinetic parameters showed altered trunk joint moments in CLBP individuals during gait. There was a significantly decrease in thoracic and lumbar flexor joint moments but an increase in the thoracic axial joint moment in CLBP group, compared to controls. We speculate that these decreased flexor joint moments of the trunk would impose a lower loading acting on the lumbar and thoracic spine, which is probably a compensatory response to protect the painful area and improve spinal stability during gait (Shum et al., 2007). The increased thoracic axial joint moment may also be considered an adaptive mechanism to compensate for the demonstrated decreased joint moment in other plane (sagittal) of the same segment. This may be representative of altered motion coordination and may lead to asymmetrical loading in spinal tissues (Shum et al., 2007). Both sit-to-stand and forward/backward bending activities were examined in a sample of sub-acute LBP patients and similar adaptations regarding trunk kinetics were found (Shum et al., 2007, 2010). Therefore, we can hypothesise this may be indicative of a common protective pattern, regardless of the functional activity (sit-to-stand, forward/backward bending or gait), that is maintained from sub-acute to chronic state.

The aforementioned motor adaptations of CLBP individuals to pain may be beneficial in the short term, however, may have consequences that could lead to further problems in the long term (Hodges & Moseley, 2003). According to Hodges and Tucker (2011), if

movements are performed in an optimal or efficient manner in a non-pain state, departure from this state due to increased or modified load, decreased movement, decreased variability, or other changes, may not be ideal. Some variability in the performance of movement has the advantage of varying the areas of joint load, muscle activity, and ligament stress. The maintenance of altered patterns of trunk variability and kinetics over time could lead to negative adaptations in CLBP patients (Hodges & Tucker, 2011). The clarification of whether non-resolution of adaptation patterns is associated with long-term consequences of CLBP cannot be confirmed with the current cross sectional study. A clarification through longitudinal studies is required and extremely important, since it might interfere with the aims of rehabilitation programmes in restoring the initial pattern or promoting the adaptive ones. However, independently of the pattern, eliminating the pain might not, per se, be effective in improving the kinematics and kinetics of the trunk and hips in CLBP patients (Shum et al., 2007), which means that retraining activities and movement strategies are always important (Kindermans et al., 2011).

Strengths and limitations:

This study contributed to a greater detail when looking to motion patterns of CLBP individuals, adding new information about kinetics to previous studies that focused on the variability of the trunk during gait. We interpreted the results about joint moment parameters taking into account the SEM values, which is very important to validate the results of groups' comparison against the known measurement error. There are also some limitations that require attention. In absence of a primary biomechanical outcome, we did not calculate sample size and chose a convenience sample. In order to compensate for this we based on the most usual sample sizes. Our participants were probably less symptomatic than the majority of the previously reported, because their mean pain intensity was 2.9 compared with 2.9 (van den Hoorn et al., 2012), 3.7 (Vogt et al., 2001) and 5.6 (Lamoth, Meijer, Daffertshofer, Wuisman, & Beek, 2006). However, all of them experienced CLBP according to the established criteria, which reinforces the idea that our findings may be indicative of the adaptive kinematic and kinetic changes experienced by people with CLBP. We characterized participants' physical activity level with the BPAQ because it was used in different studies with CLBP patients and was validated to Portuguese population. However, we think it might not be the suitable option to these patients since is mainly focused on sports and less on physical activities.

4.5 Conclusions

Stride-to-stride variability of lumbar and thoracic segments is significantly and inversely related in CLBP individuals, compared to controls: in CLBP group, greater variability in sagittal and transverse planes of lumbar segment was correlated with lower variability in sagittal and frontal planes of the thoracic segment (and vice versa), while in the control group, when significant, higher variability in one segment was mostly correlated with higher variability in the other. In CLBP individuals, a decrease in the lumbar and thoracic flexor joint moments but an increase in the thoracic axial joint moment is present during gait. The kinetic differences between CLBP and healthy individuals are out of the established SEM values, indicating that trunk joint moments results are not masked by the measurement error. These kinematic and kinetic results reinforce the argument that CLBP patients exhibit a protective movement strategy.

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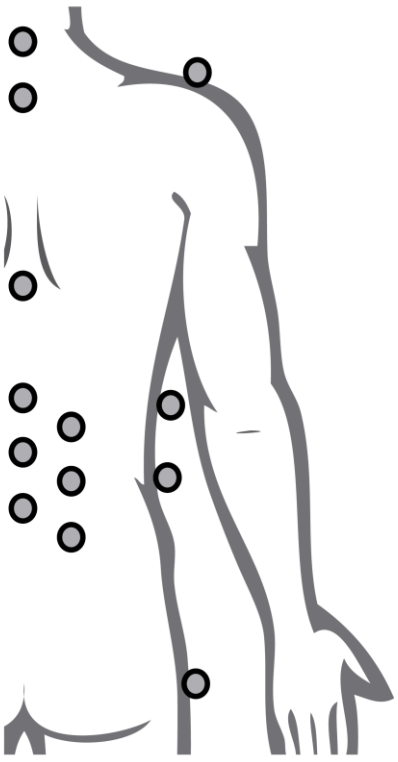
5

General Discussion



5. General Discussion

The main purpose of this thesis was to gain better understanding about the differences in the biomechanics of the trunk and lower limbs during gait in CLBP and healthy individuals. Without the detail of the separate discussions of each study (Chapters 2 to 4), in this chapter the main results of this thesis are overviewed and discussed. Methodological issues are considered and recommendations for future research are formulated.

In daily practice, clinicians often observe and treat physical impairments and movement pattern deviations of CLBP individuals, such as increased trunk stiffness (Hodges, Hoorn, Dawson, & Cholewicki, 2009; Van Daele et al., 2010), poor proprioception (Descarreaux, Blouin, & Teasdale, 2005), and altered patterns of activation of abdominal (Silfies, Mehta, Smith, & Karduna, 2009) and extensor muscles (Hides, Jull, & Richardson, 2001; Hides, Gilmore, Stanton, & Bohlscheid, 2008). Changes in the mechanical behaviour of these patients seem to be consequence of adaptive responses that result in higher trunk stiffness and modified load distribution. In the short-term, these changes may be beneficial as they protect the painful area. However, in the long-term, they might translate into poorer movement, lower variability and abnormal distribution of the load pattern (Hodges & Tucker, 2011), compromising function. We can speculate that mechanical adaptations of CLBP individuals are of clinical importance and that clinicians should assess them. To characterise differences between movement patterns of patients and healthy individuals, several comparative studies have been performed. These studies report several altered variables in CLBP patients, namely joint angles, trunk-pelvis coordination or gait speed. However, these studies lack information on reliability and measurement faults and do not give reliable extensive information to the clinicians in what manner CLBP patients adapt their movements and kinetics. So, to achieve a deep understanding of real differences between CLBP patients and healthy subjects, clinicians should be able to build their clinical reasoning on reliable 3DGA analysis, in which main movement patterns in healthy and CLBP individuals are identified. These movement patterns should take into account how large the variance of a given variable is in per group. Knowledge about these “real” differences is of importance, since it may contribute to the development of a reliable and valid assessment of CLBP individuals. Even more, it might challenge the clinician to adapt the motor control training during gait to the specific needs of an individual CLBP patient. Hence, to identify the appropriate outcomes in kinematics and kinetics as “main gait movement patterns” (Chapter 4) and to reliably compare CLBP

and healthy individuals taken into account the variances, two reliability studies were first conducted (Chapters 2 and 3).

5.1 Main Findings

Considerable evidence, available from the studies presented in chapters 2 and 3 of this thesis, supports the idea that measurement results obtained using 3DGA are reproducible in test-retest situations. In the first study (Chapter 2) test-retest reliability and MDC of 3DGA in a sample of CLBP patients was investigated. To our best knowledge this was the first study that evaluated test-retest reliability and MDC of 3DGA in CLBP patients. The results indicate high test-retest reliability for lower limb and trunk kinematics during gait in CLBP individuals, together with a clinically acceptable measurement error. Specifically, most of the joint angle peaks show ICCs greater than 0.80 (95% CI 0.36 to 0.99), while the remaining showed ICCs between 0.70 and 0.80 (varying from 0.53 to 0.97). The 95% LOA intervals for joint angles are relatively wide, pointing out that a substantial difference in an individual joint angle would be required to allow us to confidently state that a real individual change had taken place (Mieritz, Bronfort, Jakobsen, Aagaard, & Hartvigsen, 2014). Therefore, these measurements may be particularly useful for detecting change in groups and more appropriate for research purposes. Wide 95% LOA interfere with the interpretation of the results of two consecutive evaluations when individual changes occurred, being less recommended to assess individual changes. The absolute value of the SEM was small (≤ 2.5º) for the majority of joint angles and absolute MDC values were generally higher in transverse plane parameters (particularly in trunk and hip). The absence of data on test-retest reliability and MDC of 3DGA in CLBP patients precludes comparisons with standard data regarding this population. Nevertheless, a systematic review (McGinley, Baker, Wolfe, & Morris, 2009) examining the reliability of three-dimensional kinematic gait measurements in healthy individuals and in individuals with pathology, such as stroke or cerebral palsy, reported error values between 2º and 5º. With the exception of the hip, lumbar and thoracic transverse plane parameters, the error values of our study fell between 1º and 3º, which is in line with the mentioned study. Accordingly to McGinley et al. (2009), errors between 2º and 5º are likely to be regarded as reasonable although may require consideration in data interpretation, which suggests that all of our kinematic parameters results have an acceptable clinical level of error and its use can be considered. As previously mentioned, CLBP individuals have altered trunk, pelvis and hip movement patterns. Although conflicting, literature has shown that differences are mainly in the axial

rotation of the pelvis (Müller, Ertelt, & Blickhan, 2015), lumbar (Gombatto et al., 2015) or thorax (Crosbie, de Faria Negrão Filho, Nascimento, & Ferreira, 2013), as part of an adaptive response to allow a short-term protection from further pain, injury, or both. The results of this study show that hip, lumbar and thoracic rotations are the parameters with higher SEM. Caution should be taken when interpreting transverse plane differences between CLBP and healthy individuals, as they are particularly affected by measurement error. Temporal distance parameters showed excellent reliability and agreement, with ICCs varying between 0.86 (left lower limb stance time) and 0.97 (stride width). Contrary to joint angles, the 95% LOA intervals for these parameters were very restricted, indicating that these parameters are appropriate for detecting individual and group changes (Mieritz et al., 2014). Due to their excellent reproducibility, time-distance parameters are a good indicator of true stability between measurements and its use might be appropriate in clinical context.

Joint moments were less reliable than joint angles, nevertheless, most parameters still showed an ICC above 0.7, varying from 0.38 (peak of ankle internal rotation) to 0.94 (peak of ankle external rotation). A possible explanation for these less satisfactory results is that correlation coefficients might be affected by the range of variation of the parameter within the sample, with larger variations between subjects resulting in higher ICC values (Rankin & Stokes, 1998). The standard deviations of the mean values of joint moments were very low, which may be representative of their reduced variation and may have contributed to low ICCs. Despite the poorer reliability, joint moments revealed better agreement than joint angles, which is illustrated by the absolute SEM value (range between 0.01 and 0.12) and, especially, by the small SEM% (Nair, Hornby, & Behrman, 2012). Overall, the main results of this study indicate that joint angles are more suitable to distinguish CLBP patients from each other, despite the measurement error (higher reliability), while joint moment parameters are particularly appropriate for accurately assessing the scores of consecutive repeated measurements (higher agreement) (de Vet, Terwee, Knol, & Bouter, 2006).

In this study, specific methodological issues were taken into account, in order to guarantee an optimal design and enhance the generalizability of the findings. The two measurements were separated by 7 days, with all evaluations occurring at the same time of the day to diminish interference of recall bias. The assessor was blind to the results of the first assessment and the participants were allowed to adopt their normal gait (Monaghan, Delahunt, & Caulfield, 2007) and a very stable pace. Since reliability may vary across differing levels of LBP severity (Streiner & Norman, 2008) participants with different levels of pain intensity (NRS score 0 to 8) and disability (QBPDS-PT

score 0 to 41) were included. This also allows patients' level of pain to be first considered in reliability analysis and thus, its interference when comparing CLBP and healthy individuals (Chapter 4) is controlled. Therefore, we are confident to state that the results of this study offer a robust estimate of the reliability and measurement error of lower limb and trunk data during gait in CLBP.

Because knowledge about reliability and MDC values from the healthy population is extremely important in the interpretation of pathological data, the second study of this thesis (Chapter 3) aimed to investigate test-retest reliability and MDC of 3DGA kinematic and kinetic data in a sample of healthy individuals, using a two rigid segment model for the trunk. As far as we know, this was the first study that examined test-retest reliability and MDC of kinematic and kinetic 3DGA, considering thoracic and lumbar segments separately. The results suggest varied reliability levels for multi-segment trunk joint angles and joint moments, with acceptable reliability and level of error in sagittal plane. The findings reveal that lumbar sagittal kinematics is reliable, with ICCs of 0.91 (95% CI, 0.79 to 0.96) and 0.90 (95% CI, 0.77 to 0.96), and presents a measurement error around 3º. Thoracic segment exhibits ICCs between 0.66 and 0.90 in sagittal/frontal planes and good agreement, with SEM values varying between 1º and 2º. Both lumbar and thoracic segments present error values within the proposed reasonable interval (2º to 5º) (McGinley et al., 2009). However, since the variability of each parameter relative to the mean influences the interpretation of the measurement error, caution is needed when interpreting joint angles mainly based on absolute values of error. Accordingly, we looked into SEM% and verified that despite the higher ICCs in lumbar flexion/extension peaks, moderate changes are needed to indicate a real change in both parameters (43% and 33%). Interestingly, thoracic right and left lateral bending show a similar SEM% (27% and 28%), but significantly differ in ICCs (0.90 and 0.77 for thoracic right and left lateral bending). These results support the importance of considering both level of agreement and reliability parameters when interpreting reproducibility and give an indication of how precise kinematics of lumbar or thoracic segments can be measured.

Almost half of the measured lower limb joint angles (obtained from averages of 10 gait cycles) show ICCs larger than 0.80 (varying from 0.81 to 0.95), while the other half show values between 0.65 (peak of knee adduction) and 0.79 (peak of knee internal rotation). In line with McGinley et al. (2009), reliability of transverse and frontal planes was generally lower (median of 0.70 and 0.76) than of sagittal plane (median of 0.81). It is difficult to compare and discuss the results of this study with regard to published

studies due to the differences in type of studies', participants included and methodologies adopted (McGinley et al., 2009). However, our findings are consistent with literature and indicate that comparatively to transverse and frontal planes, sagittal plane joint angles are more suitable to distinguish individuals from each other, despite the measurement error (higher reliability). This also supports the importance of sagittal plane parameters when comparing joint angles between CLBP and healthy individuals.

Contrary to what was found for the CLBP group, reliability of the joint moment peaks in the healthy group was similar to the peaks of joint angles, with almost half of the parameters showing ICCs equal or above 0.80 (range between 0.80 and 0.98). These results agree with Wilken et al. (2012) and are slightly better than Meldrum et al. (2014). However, comparisons with those studies may be limited since the authors only analysed lower limb kinetics. Likewise, in the CLBP group, lower SEM% was found for joint moments comparatively to joint angle peaks, suggesting that these parameters are particularly useful to accurately assess consecutive evaluations. Although future studies focussing on the responsiveness of 3DGA are needed, we speculate based on findings of this study that kinetic parameters are more responsive and appropriate to detect minimal changes in individuals' performance over time.

The current study overcomes one of the main limitations of 3DGA reliability studies: lack of information regarding sample size calculation (McGinley et al., 2009). This is important, as small sample sizes influence the accurate representation of the population, the generalizability of the results and increase the risk of type II error (Kraemer & Thiemann, 1987). Moreover, sample size calculation increases the accuracy when interpreting the obtained results, which is one of the strengths of this study and contribution to knowledge in this area. With the chosen sample size, we guarantee the accuracy of our results and the confidence in its dissemination. We did not calculate the needed sample size for the CLBP reliability study, which is clearly a limitation when comparing results from both groups. When considering the same sample size in both groups (n=14), reliability of joint moment parameters was similar, which proves the influence of the sample size in the results. Interestingly, when compared to healthy individuals (n=23), patients (n=14) show better reliability in joint angles parameters, which may be related with higher movement variability in the healthy group.

Based on the results of the two previous studies, we conclude that both kinematics as kinetics can be used to accurately compare individuals with and without CLBP. This holds especially for the sagittal plane joint angles and joint moments, as well as time

distance-parameters. The third study of this thesis aimed to: 1) determine differences in thoracic, lumbar and hip kinematics and kinetics between CLBP patients and healthy individuals during gait, taking into account the error values; 2) and to gain insight into the variability of movement between thoracic, lumbar and hip segments in association with joint moments, in CLBP patients versus healthy individuals. To our best knowledge, this was the first study comparing multi-trunk kinematics and kinetics during gait in these two groups and our results suggest that stride-to-stride variability of lumbar and thoracic segments is significantly and inversely related in CLBP individuals, compared to controls. Specifically, in CLBP group, the sagittal and transverse planes residual rotations of the lumbar were positively correlated in magnitude and negatively correlated in sign with sagittal and frontal planes residual rotations of thoracic segment, while in the control group, when significant, both trunk segments were mostly positively correlated in sign and magnitude. This indicates that higher stride-to-stride variability in sagittal and transverse planes of lumbar segment was correlated with lower variability in sagittal and frontal planes of the thoracic segment in CLBP group (and vice versa). Literature has consistently pointed out that stiffness of the trunk is increased in these patients during gait, which seems to be related with the reduced variability of trunk motion (Lamoth, Daffertshofer, Meijer, & Beek, 2006; van den Hoorn, Bruijn, Meijer, Hodges, & van Dieën, 2012). These findings suggest a protective strategy of LBP individuals to avoid painful motion in order to prevent recurrence or pain provocation (van Dieën, Selen, & Cholewicki, 2003). This is maybe an adaptive response in acute pain but limits function when pain is maintained over time. Our results also appear to indicate that adjacent trunk segments may show different stride-to-stride variability during gait in CLBP individuals, which may suggest that increasing stiffness in one part of a system does not always increase the overall system's stiffness (Reeves, Narendra, & Cholewicki, 2007). Although literature has strongly pointed out that CLBP individuals tend to move their lumbar and pelvic segments as a rigid unit and present a less flexible pelvis-thorax coordination (Lamoth et al., 2002), the significant differences in variability between two adjacent trunk segments have not been previously reported.

The results of this study also showed that there is a decrease in the lumbar and thoracic flexor joint moments, but an increase in the thoracic axial joint moment during gait, in CLBP individuals. These differences are out of the established SEM values (Fernandes, Armada-da-Silva, Pool-Goudzwaard, Moniz-Pereira, & Veloso, 2015, 2016), indicating that trunk joint moments results are not masked by the measurement error. We speculate that these decreased flexor joint moments of the trunk would

impose a lower load acting on the lumbar and thoracic spine, which might be a compensatory response to protect the painful area and improve spinal stability during gait (Shum, Crosbie, & Lee, 2007). Although we expect this to be a typical response in the acute phase of LBP, maybe it will turn into learned behaviour by a positive reaction as less pain during this protective reaction. The increased thoracic axial joint moment may also be considered an adaptive mechanism to compensate for the demonstrated decreased joint moment in other plane (sagittal) of the same segment. These responses may be representative of altered motion coordination and may lead to asymmetrical loading in spinal tissues (Shum et al., 2007). This altered kinetic response has already been described in the literature regarding the performance of other functional activities by CLBP patients (e.g. sit-to-stand, forward/backward bending) (Shum et al., 2007; Shum, Crosbie, & Lee, 2010). Therefore, we hypothesise that this may be indicative of a common protective pattern or adaptive response in LBP, regardless of the functional activity (sit-to-stand, forward/backward bending or gait), that maintains from sub-acute to chronic state. This pattern should be assessed in LBP patients by clinicians, to indicate if adaptive responses are present.

This study contributed to a greater knowledge about the causes of motion patterns in CLBP individuals, by adding new information about kinetics to previous studies that focused on the variability of the trunk during gait. One limitation of the prior work is that the whole trunk has been considered a single rigid segment. Using a two rigid segment model for the trunk added valuable information to existing literature. Nevertheless, future studies should consider a more detailed analysis of the different parts of the trunk by using multi-trunk segments models. Another contribution of this study is that we interpreted the results about joint moment parameters taking into account the SEM values, which is very important to validate the results of groups' comparison against the known measurement error.

5.2 Methodological Considerations

Although the materials and methods used to carry out the studies included in this thesis are described with detail in chapters 2 to 4, there are still some noteworthy methodological considerations that will be addressed in the following paragraphs.

Three-dimensional model choice

Trunk motions have been tracked and described using several different models. Full 3D rotations between a few rigid body segments have been reported (Crosbie et al., 2013; Gombatto et al., 2015; Lamoth et al., 2002, 2006; Leardini, Biagi, Merlo, Belvedere, & Benedetti, 2011), as well as simpler two-dimensional (2D) projection

angles (Ferrarin, Rizzone, Lopiano, Recalcati, & Pedotti, 2004; Ferrarin et al., 2002; Frigo, Carabalona, Dalla Mura, & Negrini, 2003; Gutierrez, Bartonek, Haglund-Åkerlind, & Saraste, 2003). The trunk was demonstrated to play a fundamental role in gait, being not just a 'passenger unit', but also an active segment (Cappozzo, 1983; Kavanagh, Barrett, & Morrison, 2006). This anatomical complex was demonstrated to contribute considerably to the achievement and control of locomotion (Cappozzo, 1983; Kavanagh et al., 2006), particularly by regulating oscillations in the three anatomical directions (Kavanagh et al., 2006; Wu et al., 2004). Considering the contribution of the trunk in locomotion and the fact that in this study we aim to compare CLBP and healthy individuals during gait, looking in detail to this segment is crucial. According to the literature presented in the previous chapters, trunk axial rotations are particularly altered in CLBP patients, which supports the use of a model that measures transverse plane rotations. Therefore, choosing a 3D rigid segment model over a 2D appeared to be the most suitable option.

Two-segment model choice

Representation of overall motion of the trunk or of relevant subparts is very complex, because it occurs at many different small joints (Leardini, Biagi, Belvedere, & Benedetti, 2009). Most of the single bones of the trunk (for example, ribs and vertebrae) are hard to be tracked in 3D (Konz et al., 2006) and therefore, overall motion resulting from the aggregation of individual bones is usually reported. This accounts for the very simple models for the trunk, frequently assumed as a single rigid body, implemented in the majority of the current protocols for gait analysis (Leardini et al., 2009). Current models for trunk motion tracking have been used to study specific clinical problems (Ferrarin et al., 2002; Gupta, Vankoski, Novak, & Dias, 2005; Romkes et al., 2007), namely LBP patients' mobility impairments (Seay et al., 2011). The main difference between these models concerns the individual segments analysed: thorax or the entire trunk all together (Chung, Park, Lee, Kong, & Lee, 2010; Ferrarin et al., 2002; Nguyen & Baker, 2004), spine and shoulder line (Frigo et al., 2003), lower trunk and lumbar (Crosbie et al., 2013), as well as upper and lower lumbar (Gombatto et al., 2015). In order to have a better understanding about the gait limitations presented by CLBP individuals, this thesis focuses not only in patients' trunk tracking and in trunk-pelvis interaction, but also in the relation between the different parts of this segment, namely upper and lower trunk. With that in mind, we combined two models from two different studies, allowing the description of kinematic and kinetic data of the trunk using a two rigid segment model (Leardini et al., 2011; Seay, Selbie, & Hamill, 2008).

Marker set choice and segment coordinate systems definition

Many alternative trunk models are described in the literature with different levels of complexity. Most of them adopt very different marker sets, using mostly single markers but also rigid clusters (Konz et al., 2006; Krebs, Wong, Jevsevar, Riley, & Hodge, 1992; Wu et al., 2004). This has produced a variety of technical and anatomical coordinate systems and joint conventions, resulting in confusing terminology (for example trunk rotation in the sagittal plane has been indicated as flexion–extension or simply flexion, tilt, forward inclination, pitch or sagittal inclination) (Leardini et al., 2011).

The two rigid segment model of the trunk adopted for this study comprises the lumbar and thoracic segments. Similarly to other segments, kinematic modelling of the lumbar spine as a rigid segment can involve the placement of a set of skin surface markers over the lumbar region at specifically defined points (Crosbie, Vachalathiti, & Smith, 1997; Seay et al., 2008) or alternatively a rigid cluster, placed at a specific anatomical location on the lumbar spine (Needham, Naemi, & Chockalingam, 2014; Schache et al., 2002; Whittle & Levine, 1999). Due to the absence of recommendations from scientific organizations (for example, International Society of Biomechanics) regarding specific standards for rigid lumbar segment kinematics (only standards concerning intervertebral motion between adjacent vertebrae are reported) (Wu et al., 2002), most protocols are designed based on clinical or research specific problems and often involve different marker sets. This fact limits the reproducibility of the procedures (Needham, Stebbins, & Chockalingam, 2016). Based on Seay et al., (2008), in this thesis we modelled the lumbar spine as a rigid segment and used a protocol with markers placed across the lumbar region. To accomplish a complete trunk model, the upper trunk was also modelled with a marker set based on a previous study (Leardini et al., 2011). By adding lower limb segments, whose marker set were based on the calibrated anatomical system technique (Cappozzo, Catani, Della Croce, & Leardini, 1995), a 9-segment model (feet, shanks and thighs) was used in this thesis.

Thirty-seven passive markers and four marker clusters (thighs and shanks) were used (Figure 11).

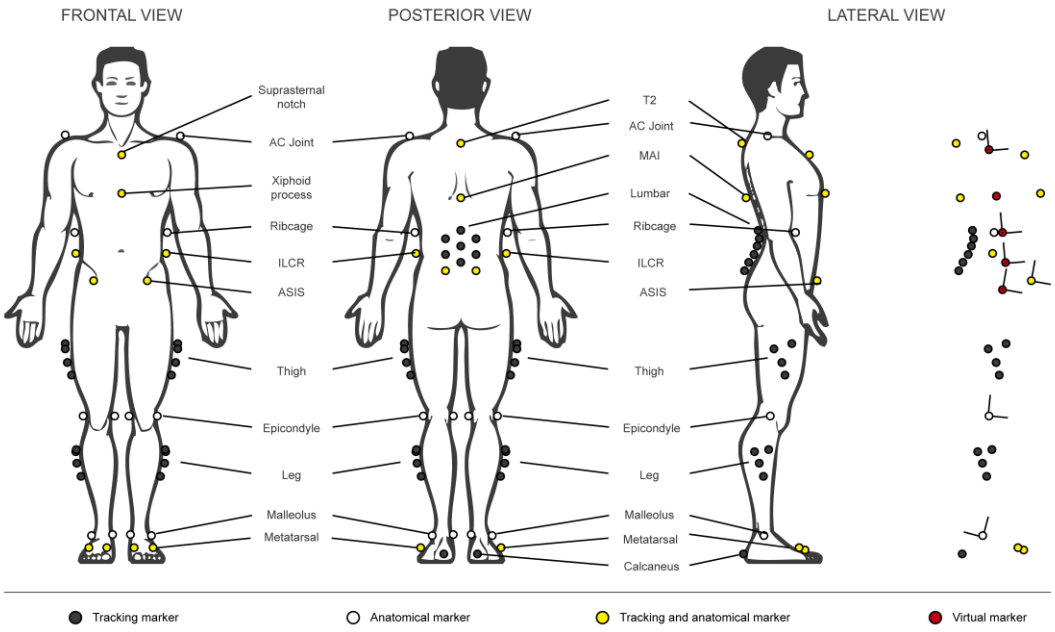


Figure 11 - Market Set and Segment Coordinate Systems

In the upper trunk six markers were attached: one on top of the spinous process of the second thoracic vertebra (T2), one over the suprasternal notch, one over the xiphoid process, one in the top of each acromioclavicular joint, and one on the midpoint between the inferior angles of most caudal points of the two scapula (MAI) (Leardini et al., 2011). The proximal end of the segment was defined using a virtual marker created in the midpoint between suprasternal notch and T2 markers. For the distal end, a virtual marker was projected in the line formed by the two bilateral ribcage markers using the previous mentioned virtual marker (proximal end), together with the midpoint between xiphoid process and the inferior angles of the scapulae. The longitudinal axis of the thorax was then established according with these two points. This axis together with the xiphoid process marker, were used to define the segment's sagittal plane and the medial-lateral axis was then determined using the cross product. Finally, the anterior posterior axis was determined by the cross product between the longitudinal and the medial-lateral axis. The markers on top of the suprasternal notch, the xiphoid process, on the MAI and on the T12-L1 joint space were used as tracking markers.

At the lumbar, individual markers were placed on the T12-L1 joint space, on the L5-S1 joint space and midway between these both markers (Seay et al., 2008). Four markers were placed over the lumbar region, on each side of the midline markers, with a minimum distance of 4 cm between all markers. Four markers were placed bilaterally

on the iliac crest and bilaterally on the ribcage at the level of the T12-L1 intervertebral joint space, immediately superior to the iliac crest markers (Seay et al., 2008). The proximal end of the lumbar segment was the previously described virtual point defined as the distal end of the thoracic segment. In order to establish the distal end of the lumbar segment, first a virtual point was created at 5% of the length of a virtual line that ran from the L5-S1 marker to the midpoint between the two anterior superior iliac spine markers. Then, another virtual marker (distal end) was projected from the previous point in the thorax longitudinal axis. The SCS was defined using the marker on top of the L5-S1 joint space and the lumbar longitudinal axis in the similar manner that was previously explained for the thorax. All the markers placed on top of the lumbar region, as well as the iliac crest markers, were used as tracking markers.

At the pelvis, two markers were placed on each anterior and posterior superior iliac spine, as well as on top of each iliac crest. The proximal end was defined using the virtual marker created as the distal end of the lumbar segment and the distal end was defined as the midpoint between the hips. The hip joint centres were computed using the pelvis markers and according to published regression equations (Bell, Pedersen, & Brand, 1990). Markers were also placed bilaterally on both medial and lateral femoral condyles, medial and lateral malleoli and first and fifth metatarsal heads to define the proximal and distal ends of the thighs, shanks and feet. Specifically the hip joint centres were defined as the proximal end of the thighs, the midpoint of the femur epicondyles as the knee proximal end and the midpoint of the tibia malleoli as the ankle proximal end (Robertson et al., 2014). The pelvis SCS, as well as all the other lower limb SCS, were defined in accordance with Robertson el al. (2014). A second LCS (for kinematic computations only) was created based on the CODA pelvis model (Robertson et al., 2014) in order to achieve a more clinically recognisable pelvic tilt (sagittal plane). Rigid marker clusters were attached to both thighs and shanks and were used to track these segments. In the feet one extra marker was placed at the most posterior aspect of the calcaneus, which together with the metatarsals heads, were the tracking markers for these segments.

POSE estimation algorithm choice

The use of video-based stereophotogrammetry in human movement analysis requires determination of the poses (position and orientation) of the body segments from skin-mounted markers before their kinematics and kinetics can be calculated (Lu & O'Connor, 1999). The principal assumption of the three pose algorithms (direct, segment and global optimization methods) is that markers move rigidly with the body

segments to which they are attached (Robertson et al., 2014). It is accepted, however, that markers attached to the skin move relatively to the underlying skeleton – STA, and that motion-capture markers produce data that may be noisy, distorted or missing (Cappozzo, Catani, Leardini, Benedetti, & Croce, 1996). Marker noise and STA result in poor estimations of pose and thus, minimizing the effect of this noise through marker placement and choosing an appropriate estimation algorithm improves the estimation of the pose (Robertson et al., 2014). In this thesis, thoracic and lumbar segments are represented as rigid segments. However, we know that these skeletal structures are not true rigid segments and thus, using a global optimization method, especially if physically realistic joint translations constraints are added to the model, may introduce higher errors. Taking that in consideration, segment optimization method was chosen and each segment was considered to be independent and to have 6 degrees of freedom (Cappello, La Palombara, & Leardini, 1996).

Filtering choices

In this thesis, kinematic and kinetic noise was minimised by the use of a Woltring generalized cross-validatory cubic spline smoothing routine (GCVSPL) (Woltring, 1986). This filtering technique is often used and, when compared to others, demonstrates acceptable results (Giakas & Baltzopoulos, 1997). When using this data-smoothing technique, the user is required to specify its mean standard error (MSE). Higher MSE values result in a greater level of filtering, potentially leading to a distortion of gait parameters in phase and magnitude (Molloy, Salazar-Torres, Kerr, McDowell, & Cosgrove, 2008). A brief experiment was conducted to determine the optimal filter option for our data. We used data from one participant and tried different types of filters often used in gait analysis studies and different filter parameters, namely a fourth order Butterworth low pass filter (at 10Hz and 6Hz) and a Woltring GCVSPL routine. After visual inspection, we concluded that the Woltring GCVSPL routine (cubic spline order and mean standard error of 1 mm) affected the artifacts frequencies without changing data, particularly in the thoracic and lumbar segments. We also verified that the higher the MSE value (e.g. 0.00001), the greater the diminution of the peak values. For those reasons we chose a Woltring GCVSPL routine with an error variance of 0.0001 (MSE of 1 mm).

Other methodological choices regarding data collection and processing

It is recognized that kinematic and kinetic data of a complete gait cycle is highly informative. However, some authors consider that extracted key points from those

curves, such as a peak value or a range (Monaghan et al., 2007; Yavuzer, Oken, Elhan, & Stam, 2008), are more meaningful, as they are easier to compare and interpret than complete curves, and tend to include the most clinically relevant features of the curves (Redekop, Andrysek, & Wright, 2008). In order to have a detailed but interpretable analysis of data, this thesis combines information from entire kinematic and kinetic curves with the respective peaks.

Gait patterns of CLBP patients have been studied using different experimental protocols. In some studies the individuals walk in treadmill (Lamoth et al., 2002, 2006; van den Hoorn et al., 2012), while in others they are asked to walk on the ground (Crosbie et al., 2013; Gombatto et al., 2015). Treadmill enables the collection of many cycles in a short period of time, which is very advantageous for research, since 3DGA is very time consuming. However, it may interfere with the natural gait pattern of individuals, suggesting that walk on the ground is more appropriate when the aim is to characterize the natural gait pattern.

Statistical decisions regarding reliability studies

A key question in the reliability of 3DGA data is whether the measures are reliable enough for clinical decision-making. The ICC is one of the most commonly reported reliability indices, however, often without mentioning which ICC formula that has been used (that is with or without inclusion of the systematic difference between measurements) (de Vet et al., 2006). Accordingly to Shrout & Fleiss (1979) there are 3 ICC models. Model 1 (one-way random effects) is used when each subject is assessed by a different set of randomly selected raters, model 2 (two-way random effects) is used when each subject is assessed by each rater and raters have been randomly selected and model 3 (two-way mixed effects) is used when each subject is assessed by each rater but the raters are the only raters of interest (in the case of this study, 1 rater). The numerical values computed with ICC models 2 and 3 are similar, although the interpretations and assumptions (as described) are different. In this study the same rater (which was the only rater of interest) assessed each subject, so the two-way mixed effects (ICC3,k) was the suitable model.

It is now well-recognized that, in isolation, correlation indices do not tell us whether the measures are ‘reliable enough’, with even high values potentially hiding measurement errors judged to be of clinical importance (Luiz & Szklo, 2005). Furthermore, expressing data variability as a coefficient results in units that are difficult to interpret clinically (Leardini et al., 2007). To be most useful, variability should be expressed in a manner that can be directly related to the measurement itself and in the same

measurement units (de Vet et al., 2006). Providing reliability parameters and no agreement parameters usually leads to wrong conclusions. It is recommended that studies reporting reliability of 3DGA data include absolute and relative measures of measurement error, such as the SEM and SEM%. There are different methods to calculate SEM and some of them are derived from the ICC formula (de Vet et al., 2006). In these methods the choice of ICC can substantively affect the size of the SEM, especially if systematic error is present (Weir, 2005). For this reason, we used the equation SEM = $SD_{diff}/\sqrt{2}$, where SD_{diff} represents the mean differences between the two measurements. As systematic error is not included in this formula, we obtained $SEM_{consistency}$, which agrees with the chosen ICC model ($ICC_{consistency}$).

In the study 2 of this thesis, we calculated the needed sample size. It allowed us to report the results of this study with a pre-defined 5% level of significance and a power of 80%, considering a minimum reliability coefficient of 0.70 and a maximum of 0.90. This guarantees the accuracy of our results and the confidence in their dissemination. Due to research options, we did not calculate the required sample size for the first reliability study. Although good reliability and agreement was obtained, we suggest that future studies should consider this procedure, since studies with high methodological quality are needed. In the absence of a clear primary outcome regarding biomechanical parameters in studies comparing CLBP and healthy individuals, it was not possible to perform sample size calculation in the third study. We based our sample size on the common sample size used in similar studies, but future studies might consider the parameters that detected differences between both groups.

5.3 Implications for future research

In this thesis we verified a significant difference in the variability between adjacent trunk segments in CLBP and healthy individuals, and found altered trunk flexor joint moments in patients. The clarification of whether the non-resolution of these altered patterns is associated with long-term consequences of CLBP cannot be confirmed with the current cross sectional study. A clarification through longitudinal studies is required and extremely important, since it might interfere with the aims of rehabilitation programmes in restoring the initial pattern or altering adaptive mal functioning responses. This longitudinal study design allows for the test of possible associations between the altered kinematic and kinetic parameters with patients' clinical profile (namely pain severity, disability or symptom duration). A better understanding of how these assumed protective strategies are established and evolve over time will allow the

development of a valid and reliable assessment and of tailored interventions that aim to enhance variability in patients, as presented in healthy individuals. In a clinical viewpoint, this is extremely valuable since it might directly contribute to alternative treatment options when dealing with CLBP individuals.

After the investigation of reliability and MDC values, future research should focus on the responsiveness of 3DGA, in order to evaluate the capability of this method in detecting (kinematic and kinetic) changes as a result of an intervention or to distinguish individual differences in response to treatment. The interpretation of whether the error magnitudes are sufficiently low must be relative to the magnitude of an expected intervention effect size and specific to population context. Further studies are necessary in CLBP individuals to provide high quality evidence indicating whether kinematic and kinetic measures are sufficiently reliable to detect clinically important change. If a measurement tool is not sufficiently responsive, it may not provide adequate evidence of effectiveness in observational studies or clinical trials. Until we know if 3DGA is sufficiently responsive to detect change in CLBP individuals after a given intervention, it is not possible to recommend its use in clinical contexts.

Different studies that attempted to estimate mobility impairments in CLBP individuals have focused on functional activities mainly characterised by flexion and extension of the trunk (e.g. sit-to-stand and reverse, putting on a sock and backward/forward bending). However, they did not take into account the measurement error, which is of importance for providing a clear conception of the differences that are important. Therefore, future studies should compare CLBP and healthy individuals in the performance of those functional activities and interpret possible differences taking into account measurement error values.

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Thesis related outcomes

First author papers in scientific journals

Published

Fernandes, Rita; Armada-da-Silva, Paulo; Pool-Goudzwaard, Annelies; Moniz-Pereira, Vera; Veloso, António P. Test-retest reliability and minimal detectable change of three-dimensional gait analysis in chronic low back pain patients Gait & Posture. 42. 491–497. 2015.

Fernandes, Rita; Armada-da-Silva, Paulo; Pool-Goudzwaard, Annelies; Moniz-Pereira, Vera; Veloso, António P. Three-dimensional multi-segmental trunk kinematics and kinetics during gait: Test-retest reliability and minimal detectable change. Gait & Posture. 46. 18–25. 2016.

Submitted

Fernandes, Rita; Pool-Goudzwaard, Annelies; Moniz-Pereira, Vera; Armada-da-Silva, Paulo; Veloso, António P. Loss of variability and altered three-dimensional trunk and hip kinetics during gait in chronic low back pain individuals. Annals of Biomedical Engineering. June 2016.

First author abstracts in scientific journals

Fernandes, Rita; Moniz-Pereira, Vera; Veloso, António P. Armada-da-Silva, Paulo. Test–retest reliability of three-dimensional gait analysis in chronic low back pain individuals: A preliminary study. Gait & Posture. 42. S82–S83. December 2015.

Fernandes, Rita; Armada-da-Silva, Paulo; Pool-Goudzwaard, Annelies. Transversus Abdominis, Internal Oblique and External Oblique thickening change during a voluntary contraction in patients with CLBP. Journal of Biomechanics. 45. S502. July 2012.

Podium presentations

Invited speaker

Fernandes, Rita; Armada-da-Silva, Paulo; Pool-Goudzwaard, Annelies; Veloso, António P. Test-retest reliability and minimal detectable change of three-dimensional gait analysis in chronic low back pain patients and healthy individuals. 9ª Congresso Nacional de Fisioterapeutas. Estoril, Portugal. 2015.

Congresses

Fernandes, Rita; Moniz-Pereira, Vera; Veloso, António P. Armada-da-Silva, Paulo; Test-retest Reliability and Minimal Detectable Change of Three-Dimensional Gait Analysis in a Sample of Chronic Low Back Pain Patients: A Preliminary Study. 1st Clinical Movement Analysis World Conference. Rome, Italy. 2014.

Seminars

Fernandes, Rita. Test-retest Reliability of Three Dimensional Gait Analysis in a sample of Chronic Low Back Pain Individuals. Friday Evenings Research Seminars - Psychometrics and clinimetrics: Two faces of the same coin? Mestrado em Fisioterapia, Especialização em Condições Músculo- Esqueléticas, da Faculdade de Ciências Médicas – Universidade Nova de Lisboa e Escola Superior de Saúde - Instituto Politécnico de Setúbal. Setúbal, Portugal. 2013

Fernandes, Rita; Armada-da-Silva, Paulo; Pool-Goudzwaard, Annelies; Biomechanical determinants of disability and pain in chronic low back pain individuals: a 3-month follow up study. Biomechanics and Functional Morphology Laboratory Seminar. Cruz Quebrada, Portugal. 2012.

Poster presentations

Fernandes, Rita; Armada-da-Silva, Paulo; Pool-Goudzwaard, Annelies; Moniz-Pereira, Vera; Veloso, António P.Test-retest reliability and minimal detectable change of three-dimensional gait analysis in chronic low back pain patients and healthy individuals. International Federation of Orthopaedic Manipulative Physical Therapy Conference. Glasgow, United Kingdom. 2016.

Fernandes, Rita; Armada-da-Silva, Paulo; Pool-Goudzwaard, Annelies. Transversus Abdominis, Internal Oblique and External Oblique thickening change during a voluntary contraction in patients with CLBP. Journal of Biomechanics. 45. S502. July 2012.

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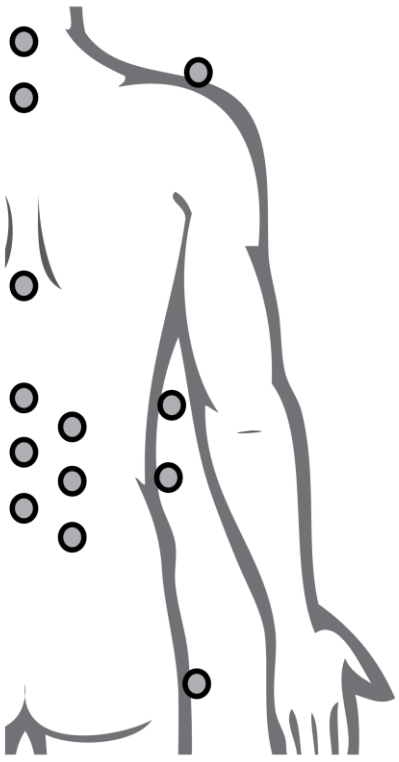
Appendix



Autorização de compilação

Eu, Paulo Alexandre Silva Armada da Silva, na qualidade de co-autor(a) do artigo "Test–retest reliability and minimal detectable change of three-dimensional gait analysis in chronic low back pain patients", publicado em 2015, no volume 42, pp.491-7 do jornal científico Gait & Posture, declaro que autorizo a inclusão do mesmo na dissertação de doutoramento intitulada "Biomechanics of the trunk and lower limbs during gait in individuals with and without chronic low back pain", submetida pela candidata Rita Noélia da Silva Fernandes.

Cruz-Quebrada, 27 de junho de 2016

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Autorização de compilação

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IPS Instituto Politécnico de Setúbal
Escola Superior de Saúde